NAVAL POSTGRADUATE SCHOOL
Monterey, California

THESIS

A NEW KINEMATIC MODEL FOR THE STUDY OF THE ROLE OF THE ANTERIOR CRUCIATE LIGAMENT (ACL) IN HUMAN KNEE MOTION

by

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December, 1995

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DTIC QUALITY INSPECTED 1

REPORT DOCUMENTATION PAGE

Form Approved
OMB No. 0704-0188

Public reporting burden for this collection of information is estimated to average 1 hour per response, including the time for reviewing instructions, searching existing data sources, gathering and maintaining the data needed, and completing and reviewing the collection of information. Send comments regarding this burden estimate or any other aspect of this collection of information, including suggestions for reducing this burden, to Washington Headquarters Services, Directorate for Information Operations and Reports, 1215 Jefferson Davis Highway, Suite 1204, Arlington, VA 22202-4302, and to the Office of Management and Budget, Paperwork Reduction Project (0704-0188), Washington, DC 20503.

1. AGENCY USE ONLY *(Leave blank)*

2. REPORT DATE
December 1995

3. REPORT TYPE AND DATES COVERED
Master's Thesis

4. TITLE AND SUBTITLE
A NEW KINEMATIC MODEL FOR THE STUDY OF THE ROLE OF THE ANTERIOR CRUCIATE LIGAMENT (ACL) IN HUMAN KNEE MOTION

5. FUNDING NUMBERS

6. AUTHOR(S)
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7. PERFORMING ORGANIZATION NAME(S) AND ADDRESS(ES)
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Monterey, CA 93943-5000

8. PERFORMING ORGANIZATION REPORT NUMBER

9. SPONSORING/MONITORING AGENCY NAME(S) AND ADDRESS(ES)

10. SPONSORING/MONITORING AGENCY REPORT NUMBER

11. SUPPLEMENTARY NOTES
The views expressed in this thesis are those of the author and do not reflect the official policy or position of the Department of Defense or the U.S. Gorvernment.

12a. DISTRIBUTION/AVAILABILITY STATEMENT
Approved for public release; distribution is unlimited.

12b. DISTRIBUTION CODE

13. ABSTRACT *(Maximum 200 words)*
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14. SUBJECT TERMS
BIOMECHANICS, KNEE KINEMATICS, ANTERIOR CRUCIATE LIGAMENT

15. NUMBER OF PAGES
74

16. PRICE CODE

17. SECURITY CLASSIFICATION OF REPORT	18. SECURITY CLASSIFICATION OF THIS PAGE	19. SECURITY CLASSIFICATION OF ABSTRACT	20. LIMITATION OF ABSTRACT
Unclassified	Unclassified	Unclassified	UL

NSN 7540-01-280-5500

Standard Form 298 (Rev. 2-89)
Prescribed by ANSI Std. Z39-18
298-102

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Submitted in partial fulfillment
of the requirements for the degree of

MASTER OF SCIENCE IN MECHANICAL ENGINEERING

from the

NAVAL POSTGRADUATE SCHOOL
December 1995

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ABSTRACT

A six degree of freedom model was utilized to continuously measure the motions of loaded cadaveric human knees with unconstrained motion at the tibiofemoral joint through a range of motion from zero to 110 degrees flexion. Several conditions were studied. Loading conditions were varied to simulate the natural body forces (i.e. the normal condition) and quadriceps-deficient condition. The range of motion in which the anterior cruciate ligament (ACL) is the primary restraint to anterior tibial translation was determined. The effect of ACL insufficiency on the kinematics of the human knee was investigated by comparing the kinematics of the knee specimens in the intact state with the kinematics obtained after the ACL was surgically severed. To simplify the complex kinematics of a six degree of freedom model, the motion of the instant center of the tibiofemoral joint for each specimen was estimated using the femoral transepicondylar pin reference point. The estimated motion of the instant center of the knees in the intact state and ACL deficient state was compared to empirical observations. The importance of the motion of the instant center was then determined in pathologic knee motion. Finally, the effect of total knee replacement on kinematics was investigated.

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I. INTRODUCTION

The kinematics of the intact human knee have been the subject of numerous studies. Yet studies investigating knee kinematics have had significant limitations. These limitations include large measurement error, discontinuous data acquisition, non-physiologic loading conditions, and constraint at the femur and or the tibia which decreases the effective degrees of motions. The reason for these limitations stems from the lack of a true unconstrained model to measure and record data with respect to a six degree of freedom coordinate system in the moving and loaded human knee. Data obtained from interrupting knee motion to measure translations and rotations cannot be assumed to be the same as data obtained from continuous knee motion. Moving only the femur and holding the tibia stationary imposes kinematic constraints on the knee which can alter mechanisms contributing to knee joint motion. The complexity of the human knee has limited the manner in which research questions could be answered. Presently, no report has described the kinematics using a true six degree of freedom coordinate system of the intact knee compared to the anterior cruciate ligament deficient knee and the reconstructed knee after arthroplasty! Further, no study has been published that compares the kinematics of different surgical techniques of cruciate ligament reconstruction using continuous uninterrupted knee motion. The assumption has been that effects of cruciate ligament reconstruction on knee kinematics approach normal when the graft is placed at the cruciate isometric point and securely fixed. Presently, this hypothesis has not been mechanically proven.

In conjunction with LCDR M. DeMaio and LCDR D. Adkison, medical doctors in the Department of Orthopaedic Surgery, National Naval Medical Center, Bethesda, Maryland, the motion of the human knee was studied using a new six degree of freedom model developed in cooperation with the Naval Post Graduate School. The range of motion of an intact, anterior cruciate ligament (ACL) deficient and reconstructed knee

was studied and compared at different conditions. Further, continuous kinematics of the intact knee will be described with each knee acting as its own control for subsequent conditions. The ACL was then severed arthroscopically and the range of motion tested under similar loading conditions as the normal knee. Once the relative motion of the knee has been recorded, isolated and coupled motions for specified degrees of freedom will be discussed. The range of motion in which the ACL is the primary restraint to anterior tibial translation will be investigated using the data obtained from the six degree of freedom model. From this information, normal kinematics and pathokinematics relating the role of the ACL will be determined. This will serve to direct future efforts in the clinical diagnosis of cruciate ligament injuries as well as to improve existing total knee replacement techniques.

II. BACKGROUND

The vast majority of published research reviewed used static models which constrained knee motion to a specific plane. Commonly, the femur is fixed and translation of the tibia is recorded at fixed intervals of flexion. The result would be displayed as continuous knee motion after data interpolation [Ref 1, 2]. A number of papers [Ref. 3, 4] utilized a six degree of freedom model but did not record data under continuous loading conditions. Others have constructed an unconstrained six degree of freedom model but do not load the specimens or record data continuously.

Several muscle groups are responsible for controlling flexion and extension of the knee joint [Ref. 5]. The primary flexors are the hamstrings and the gastrosoleus while the primary extensors are the quadriceps femoris (Figure 1). The function of the cruciate ligaments (Figure 2) and shape of the condyles are closely related. Without the cruciate ligaments, knee joint kinematics could not exist in the form to be described. This raises the question of which is present first, the cruciate ligaments or the moving articular surfaces, i.e., the contact surface between the femur and tibia, and whether the knee joint can function normally after the loss or rupture of a cruciate ligament. The functional importance of the cruciate ligaments must have a very early origin in human development. Research has shown that cruciate ligaments are formed in the human embryo by the tenth week before the development of a joint space with freedom of movement. The basic contour of the femoral condyle is also apparent at this stage. Thus the subsequent motion of the articular surfaces apparently is predetermined to a large degree by the early presence of the cruciate ligaments. Numerous references stress that the cruciate ligaments perform the function of a true gear mechanism and form the nucleus of knee joint kinematics. Moreover, cruciate deficiency disintegrates the gear mechanism, and that the latter cannot be explained by the function of the surrounding ligaments and joint tissue.

An anatomically based coordinate system was used to serve as a reference to define knee joint kinematics. The three rotations and translations are described as (Figure 3):

Rotation:

a) flexion and extension about the femoral transepicondylar line (about the x-axis) (Figure 3a)

b) internal and external rotation along the axis of the tibia (about the y-axis) (Figure 3b)

c) adduction and abduction about an axis perpendicular to both the x-axis and y-axis using the conventional right-hand rule (about the z-axis) (Figure 3c)

Translation:

a) medial to lateral shift along the transepicondylar line (along the x-axis) (Figure 3a)

b) compression and distraction along the superior-inferior direction of the tibia (along the y-axis) (Figure 3b)

c) anterior-posterior drawer along the axis in the anterior-posterior direction (along the z-axis) (Figure 3c)

The ACL has a very complex anatomy and equally complex function. As the knee undergoes flexion-extension as well as internal-external and adduction-abduction rotation, the length and orientation of the ACL changes significantly [Ref. 6]. The attachments of the ACL to both the femur and the tibia allow various portions of the ligament to be relatively taut while others are lax, depending on the knee motion. At a particular flexion angle, some part of the ligament fiber bundle experiences tension while other bundles are unloaded. Additionally, the orientation of the ACL changes with flexion angle. Hence, the mechanical behavior of the ACL in tension depends on knee orientation and on the direction of applied load.

Studies show that the ACL restricts anterior translation and limits internal rotation of the tibia with respect to the femur at angles between zero and 40 degrees of flexion [Ref. 7]. The complex arrangement of ACL fiber bundles with different lengths makes uniform loading of the entire ACL very difficult. The axis of the ACL is also difficult to define uniquely because of the complexity of its anatomical structure. This makes finding the mechanical properties such as ultimate strength and Young's modulus experimentally difficult because the data becomes unrepeatable. Ultimately, the kinematics of knee motion and the mechanical properties of the cruciate ligaments become a statistical challenge. Using a six degree of freedom model increases the difficulty in determining significant statistical information because of a lack of controlled parameters.

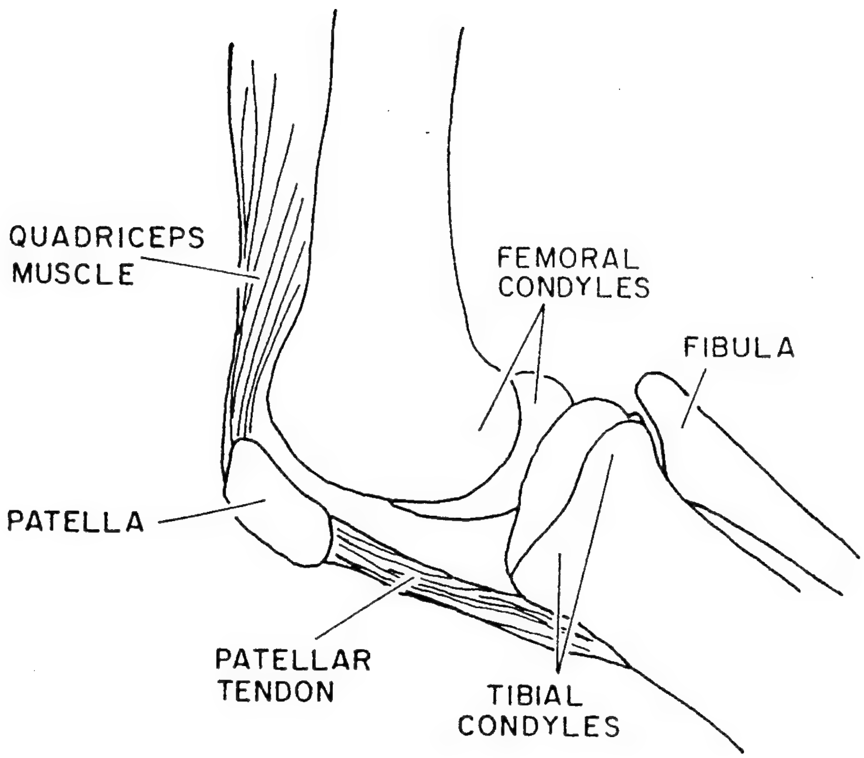

Figure 1: The Knee Joint in the Sagittal Plane Showing the Various Attachment Sites of the Gastrosoleus, Hamstrings, and Quadriceps Muscle Groups [Ref. 9].

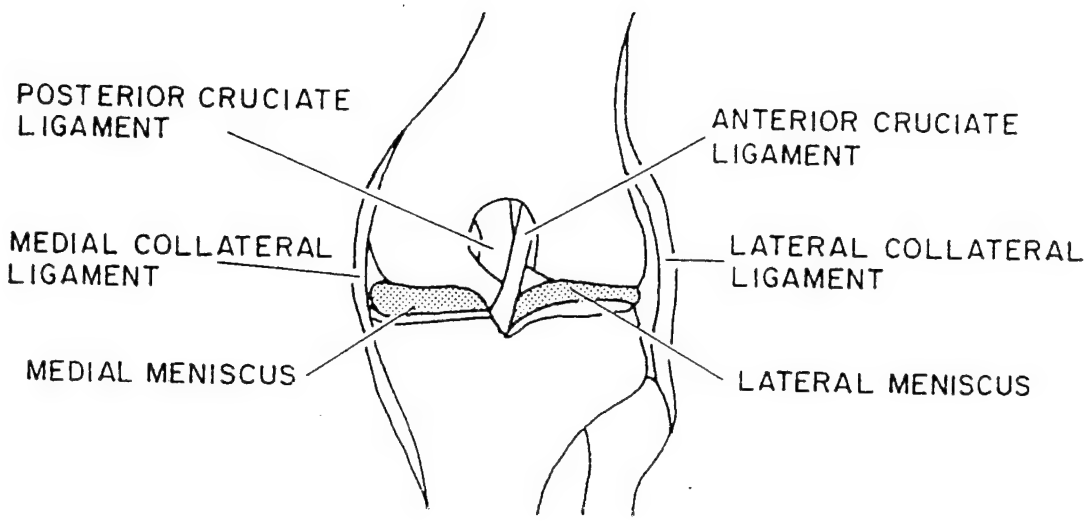

Figure 2: Anterior View of the Tibiofemoral Joint Showing the Insertion Sites of the Anterior and Posterior Cruciate Ligaments, Medial and Lateral Collateral Ligaments, and Menisci [Ref. 10].

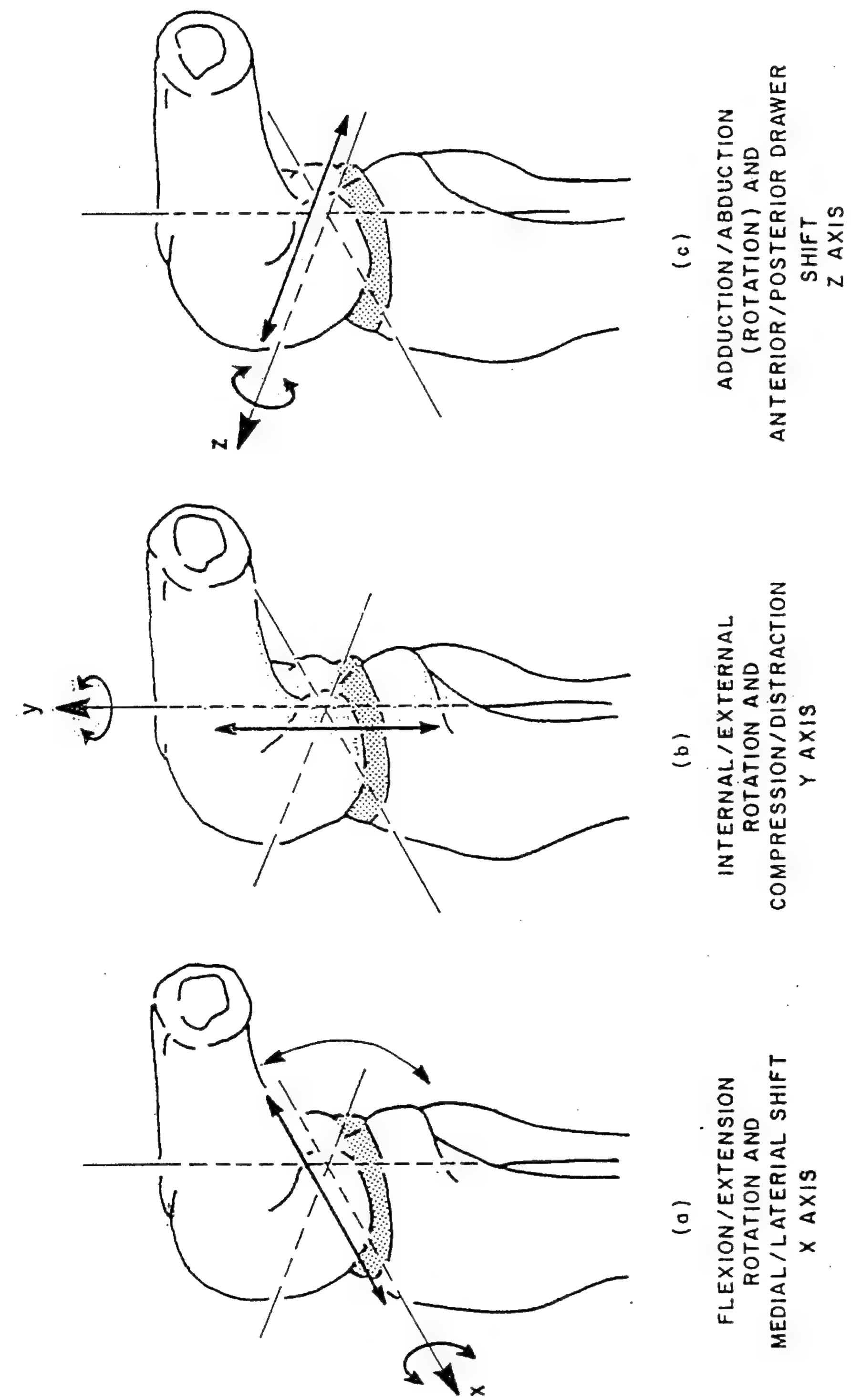

Figure 3: An Anatomical Cartesian Reference Frame for the Tibiofemoral Joint [Ref. 9, 10].

III. EXPERIMENTAL PROCEDURE AND TEST EQUIPMENT

Five right human knee specimens were used to conduct the research. The donors of the cadaveric knees were four males and one female with an average age of 52 years. The donors had no history of knee trauma. However, the donor's physical activity level and history of malalignment were unknown.

The specimens were examined and photographed arthroscopically to confirm normal cruciate ligaments and menisci. The specimens were then cut mid-tibia and mid-femur and kept frozen at 258 Kelvin. The specimens were removed from the freezer, thawed, and prepared the evening prior to the morning of the tests. Specimen preparation consisted of removal of tissue leaving the quadriceps mechanism, cruciate ligaments, and capsule intact. A 10 mm diameter Kuntschner rod (K-rod) was cemented into the medullary canal of both the tibia and femur. The rods varied in length due to the length of the bones as well as the necessity to maintain the same range of motion from zero (full extension) to 110 degrees of flexion. Eye bolts were inserted into the bones at the vector sum points for the gastrosoleus, the hamstrings, and the quadriceps muscles. The reference point for the measurements was the femoral transepicondylar line which has proven effective in studies by Hollister [Ref. 8]. Hence a transepicondylar pin was used to define the reference point and represent the center of femoral axis rotation. The instrument jig, which houses the translational transducers, was fixed anterior and below the tibial tubercle.

To obtain the six degrees of freedom desired, the femur and tibia were secured to two thin cantilevered plates which house the three orthogonal rotational transducers. The plates are attached to a "velmax unislide" rigid frame which houses the worm-drive mechanism (Figure 4). The K-rods that were inserted into the medullary canals were then inserted into a cylindrical socket which is attached to the geared y-rotation transducers. When engaged, the worm-drive mechanism moves the top plate up or down, depending

on the half-cycle run, while the bottom plate remains stationary. A strain gauge was attached to the top half of the upper plate to determine the amount of force the worm-drive motor induced at the knee joint which may polarize the kinematics of the knee specimen. This induced force was negligible compared to the simulated body forces that were continuously loaded during uninterrupted knee motion. Once the knees were anatomically prepared for testing, the specimens were mounted and loaded for the condition to be tested. Table 1 describes the three loading conditions that were used to simulate the normal, quadriceps deficient, and quadriceps loaded condition.

Condition	Quadriceps Load (lbf)	Hamstring Load (lbf)	Gastrosoleus Load (lbf)
Normal	20	10	10
Quadriceps Deficient	10	10	10
Quadriceps Loaded	40	10	10

Table 1. Specific Loading Conditions Used to Simulate the Natural Body Forces at the Tibiofemoral Joint.

The loads were attached using ropes and pulleys (Figure 5, 6) through the vector sum points of the various muscle groups to simulate loading of the knee joint. The speed at which the velmax unislide flexed or extended the knee specimens was controlled by the velmax "controller/driver". The driver maintained the speed of the worm-drive mechanism such that the specimens flexed or extended at a rate of 2 degrees per second. A half-cycle run lasted approximately one minute. For each half-cycle run, the data acquisition system recorded approximately 1500 data points which was displayed in a nine column ASCII format (three translations and six rotations). In order to maintain

similar test parameters for comparison, the length of the K-rods varied to allow for the different lengths of femurs and tibias. Hence, each knee acted as its own control for subsequent conditions.

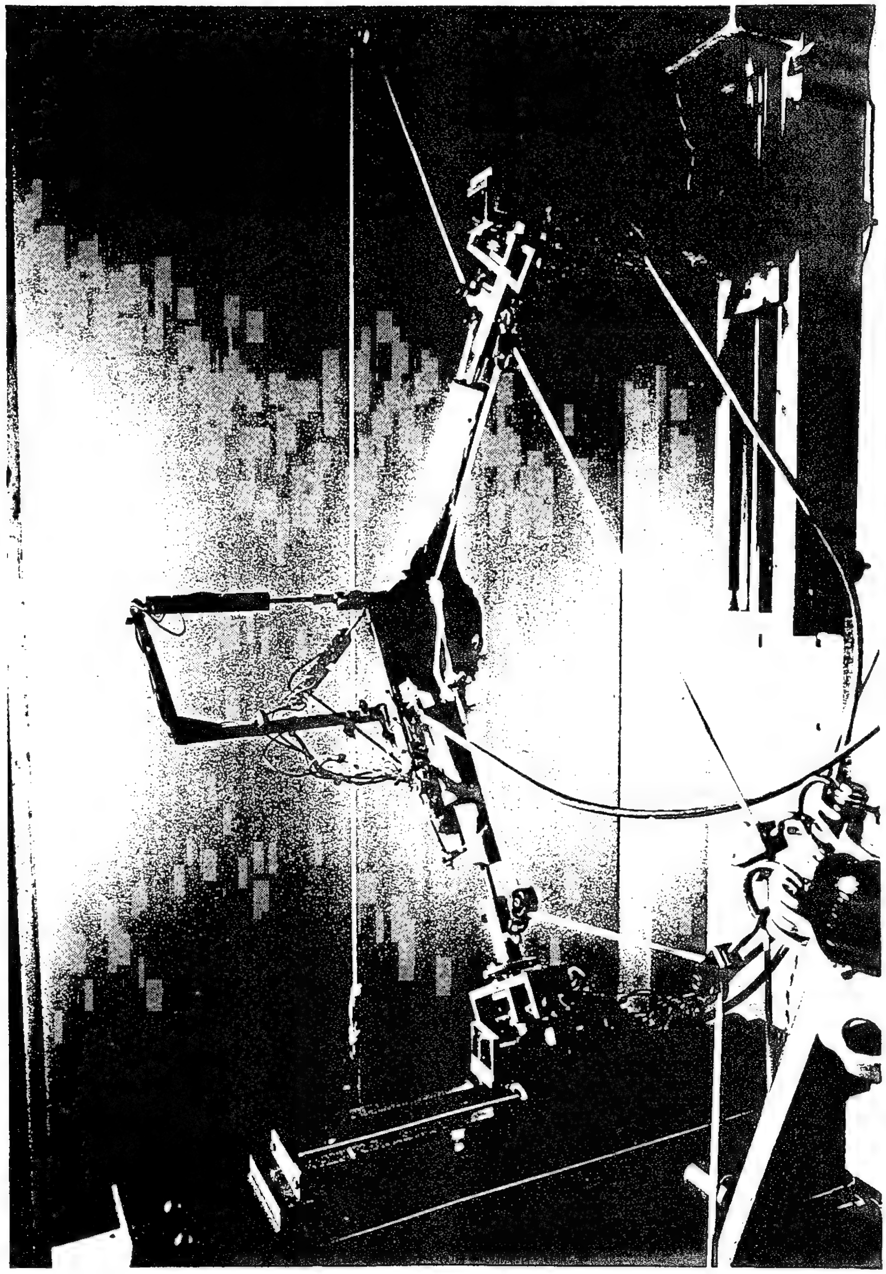

Figure 4: Photograph of the Velmax Unislide Test Stand with a Foam Model of a Knee Specimen. The Rotational Transducers are Located at the Ends of Each Long Bone Attached to the Upper and Lower Cantilevered Support Plates. The Translational Transducers are Located at the Tibiofemoral Joint.

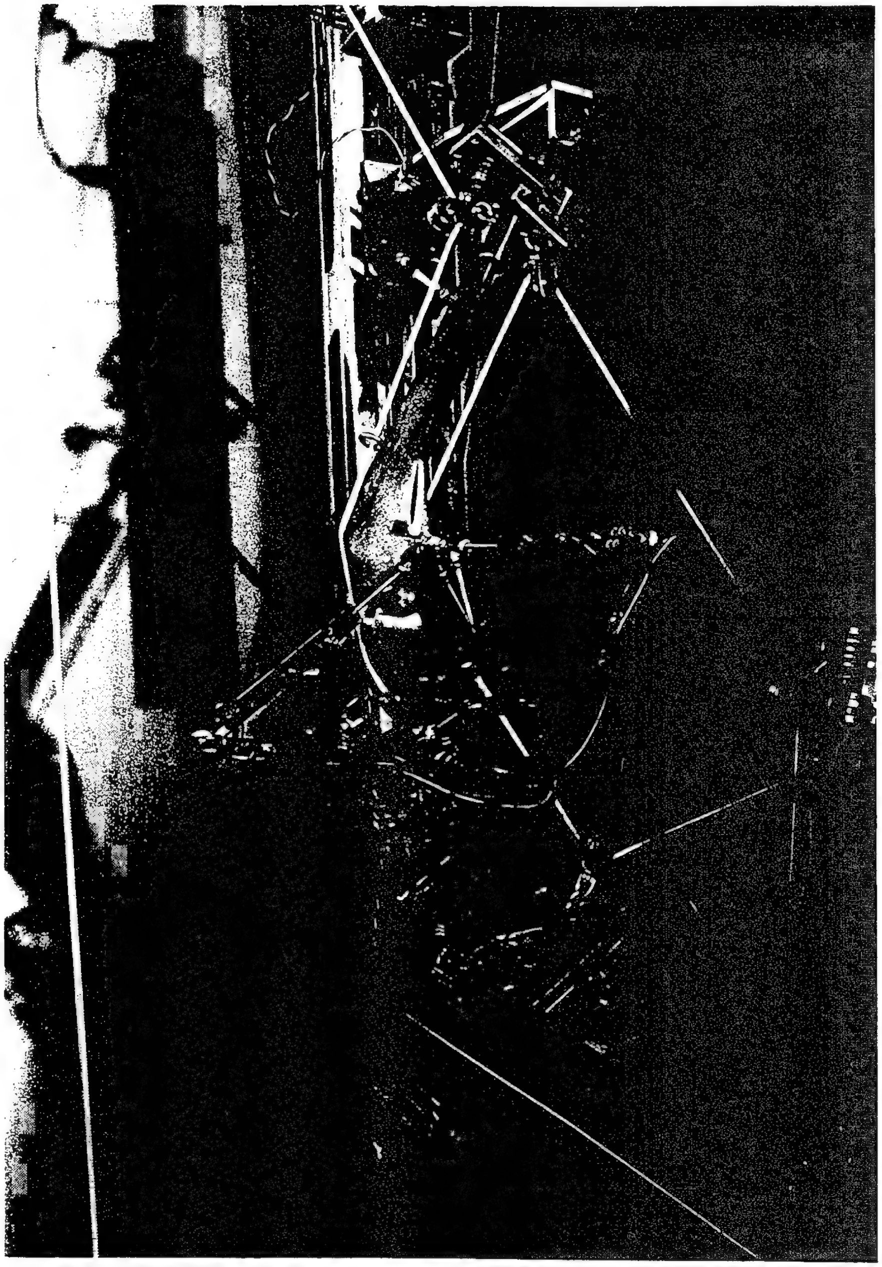

Figure 5: Photograph of the Velmax Unislide Test Stand Showing the Foam Knee Specimen in a Flexed Position. The Vertical Rope on the Left is Used to Hang the Weight Which Simulates the Quadriceps Muscle Group Force.

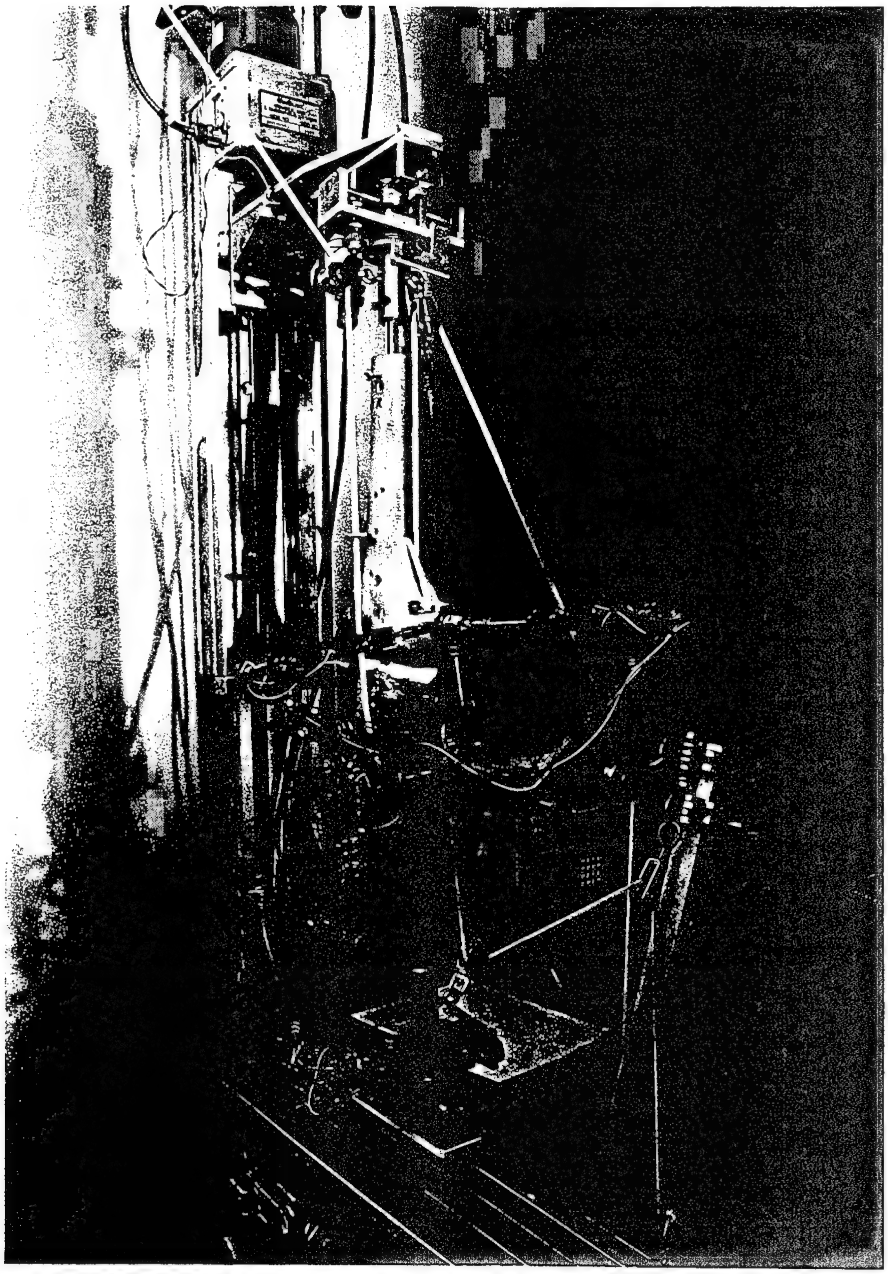

Figure 6: Photograph of the Velmax Unislide Test Stand Showing the Foam Knee Specimen in the Fully Extended Position. The Rope Located Top Right is Used to Hang the Weight Which Simulates the Hamstrings Muscle Group Force While the Rope Located Bottom Right is Used to Simulate the Gastrosoleus Force.

IV. ANALYSIS

A. DATA ACQUISITION AND REPEATABILITY

The knee specimens varied in size and shape which resulted in the amount of data acquired as well as the displacements to vary from specimen to specimen. Each knee acted as its own control for subsequent motions and no quantitative comparisons were drawn between specimens. The data acquisition system was manually controlled which resulted in different starting and stopping points along the unislide from one half-cycle of motion (extension to full flexion) to the next.

To analyze the data in terms of translations and rotations, a computer algorithm was written using the MATLAB scientific software. The algorithm converted the measured voltage data from the data acquisition system into translations and rotations using a conversion factor of 21 degrees/volt for the ungeared rotational transducers, 9 degrees/volt for the geared rotational transducers, and 3.124 mm/volt for the translational transducers. The transducers have a measurement uncertainty of +/- 0.6 degree for the rotational transducers and +/- 0.3 mm for the translational transducers with a maximum range of travel of 340 degrees and 49.53 mm, respectively. Using simple geometric relations, the six independent rotational measurements (three at the end of each long bone) were reduced to relative rotations between the femur and the tibia.

Once all of the data were converted, displacement plots were made to determine consistency among the data obtained during half-cycle runs with the same loading condition. In this way, equipment malfunction could be identified and corrected before further tests were conducted. Figures 7 and 9 shows plots of tibial translation along the anterior-posterior drawer and relative internal tibial rotation versus flexion angle, respectively, for specimens 005, 019, 037, and 221. Each curve represents consecutive half-cycle runs (extension to flexion) under the same loading condition. Figures 8 and 10 are similar plots for specimen 066. Qualitatively, the plots show that the six degree of freedom apparatus and data acquisition system performed in a highly reliable manner.

A general student's T-*test* was conducted to judge the significance between measured data of the same variable and loading condition of each knee during consecutive half-cycle runs. The general student's T-*test* tests the hypothesis that two samples have the same variances by trying to measure the significance of a difference of means. The T-value is the ratio of the difference between two means and the standard error of the difference of means from the pooled variance. The reported probability value *p*, is used to judge the significance between the two samples. The probability is a number between zero and one. A probability value close to one means that the two set of samples are significantly relevant. For each degree of freedom, there were four sets of approximately 1500 data points per set recorded for each specimen. Hence, there was approximately 30,000 measurements recorded per degree of freedom. Although not included, the general student's T-*tests* results show that the p-value of paired data for each degree of freedom with the same loading condition was consistently in the range of 0.70-0.99. Thus, the data was highly repeatable and the six degree of freedom apparatus and acquisition system performed consistently throughout the range of motion. One of the most interesting characteristics of human knee motion can be observed in Figures 11 through 15. The plots show that each knee specimen traces out a unique path of motion during flexion and extension. During the flexion phase, the tibia translates anteriorly and rotates internally relative to the femur. Contrastingly, the tibia translates posteriorly and rotates externally relative to the femur during the extension phase. Remarkably, the tibia does not follow the same path to complete either phase of motion. Similar plots can be obtained for the other degrees of freedom.

One possible explanation for this behavior lies in the primary functions of the cruciate ligaments. As previously noted, the primary function of the ACL is to act as a restraint to limit anterior tibial translation. Moreover, the range of flexion in which the ACL is most effective in limiting anterior tibial translation is between zero and 40 degrees of flexion. Hence, the plots show a few millimeters of translation in this range of flexion. Beyond 40 degrees, the PCL begins to tighten and the ACL begins to slacken. As

a result, the tibia quickly translates anteriorly without the influence of the ACL. At very large flexion angles, the ACL once again begins to tighten to limit any further anterior translation. Of course the surrounding tissue and knee joint capsule further limit any additional translation.

Contrastingly, the primary function of the PCL is to act as a restraint to limit posterior tibial translation. Moreover, the range of flexion in which the PCL is most effective in limiting posterior tibial translation is between 70 and 90 degrees of flexion. Beyond 90 degrees, the PCL begins to slacken and the surrounding tissues of the knee joint provide additional constraints to further translation. In the range of 40-70 degrees of flexion is a gray area in which both cruciate ligaments seem to be least effective in limiting their respective translations. Thus, the knee joint seems to be most vulnerable to injury in the range of 40-70 degrees of flexion. During the extension phase, the PCL seems to limit posterior tibial translation for a much wider range of flexion angle than does the ACL in limiting anterior tibial translation. This certainly is in agreement with the fact that the PCL is a larger and stronger ligament than the ACL. As the flexion angle decreases below 70 degrees, the PCL begins to slacken and allows the tibia to further translate posteriorly relative to the femur at a larger rate.

The influence of the cruciate ligaments on tibial rotation is less obvious than their influence upon tibial translation. One possible explanation is that the cruciate ligaments play a less significant role in limiting tibial rotation. Additionally, the medial and lateral collateral ligaments, knee joint capsule, and surrounding tissue may have as a primary function the responsibility of limiting tibial rotation. The cruciate ligaments would then serve as supplemental restraints to tibial rotation.

B. TRANSLATION COMPARISONS

Although translation occurs simultaneously in three planes, the translation in one plane is so great that it accounts for nearly all of the motion [Ref. 6]. Moreover, many muscles produce forces on the knee and at any particular instant one muscle group

predominates, generating a force large enough that it accounts for most of the muscle force acting on the knee. Thus, the present analysis was simplified by looking at translation in one plane and qualitatively measuring the effect that the force produced by a single muscle group has on that translation.

Figures 16 through 20 are tibial translation plots along the anterior-posterior drawer versus flexion angle of all five specimens. Each curve represents the mean translation of three half-cycle runs (extension to flexion) at a specific loading condition. The translation along the medial-lateral direction as well as compression-distraction were relatively minor in comparison and their analysis will be deferred.

Specimen 221 translated approximately 8 mm in the anterior direction at full flexion. In the range of zero to 20 degrees, the tibia translated posteriorly approximately 2 mm and beyond 20 degrees quickly translated anteriorly relative to the femur. After transection of the ACL, anterior tibial translation decreased approximately 2 mm beyond 20 degrees of flexion and paralleled the intact state. Changing the quadriceps force while in the ACL deficient condition had very little effect on tibial translation.

Specimen 037 translated approximately 18 mm in the anterior direction at full flexion. In the range of zero to 30 degrees, the tibia translated posteriorly relative to the femur for approximately 1 mm. Beyond 30 degrees, the tibia quickly translated anteriorly and achieved full translation at full flexion. After transection of the ACL, anterior tibial translation decreased by 1 mm and paralleled the intact state. Increasing the quadriceps force to 40 pounds while in the ACL deficient condition has little effect on tibial translation and decreasing the quadriceps force to 10 pounds significantly reduced anterior tibial translation beyond 40 degrees of flexion. Specimen 019 translated approximately 9 mm in the anterior direction at full flexion. In the range of zero to 20 degrees, the tibia translated posteriorly relative to the femur with a maximum value of 6 mm at 20 degrees of flexion. Between 20 and 30 degrees, there is very little translation and beyond 30 degrees, the tibia quickly translates anteriorly relative to the femur. After transection of the ACL, posterior tibial translation reached a maximum of 8 mm at 25

degrees of flexion, and beyond 25 degrees the tibia translated anteriorly at a much faster rate than the intact state. Beyond 65 degrees, the tibia translated approximately 2 mm more than the tibia in the intact state at full flexion. Increasing the quadricep force to 40 pounds decreased anterior tibial translation by 2 mm beyond 25 degrees of flexion while decreasing the quadriceps force to 10 pounds dramatically decreased the amount of anterior tibial translation beyond 40 degrees of flexion. At very large angles of flexion, the ACL deficient knee with the quadricep deficient condition became highly unstable and translated unpredictably.

Specimen 005 translated approximately 9 mm in the anterior direction at full flexion. In the range of zero to 40 degrees, there is very little change in the translation. Beyond 40 degrees, the tibia quickly translates anteriorly to its maximum value at full flexion. After transection of the ACL, anterior tibial translation increased by 3 mm and paralleled the intact state. Increasing the quadriceps force to 40 pounds while in the ACL deficient condition had very little effect on tibial translation and decreasing the quadriceps force to 10 pounds decreased anterior tibial translation below that of the other three conditions for the same flexion angle obtained.

Specimen 066 translated uncharacteristically approximately 33 mm in the anterior direction at full flexion. In the range of zero to 70 degrees, the tibia gradually translated approximately 8 mm and beyond 70 degrees the tibia quickly translated to its maximum value. After transection of the ACL, anterior tibial translation decreased significantly beyond 70 degrees of flexion. Increasing or decreasing the quadriceps force while in the ACL deficient condition had very little effect on tibial translation throughout the range of motion.

In summary, the tibia of all five specimens translated anteriorly relative to the femur at full flexion. Specimen 066 translated the farthest to reach the full 110 degrees flexion while specimen 019 translated the shortest. This trend was consistent with the observation that specimen 066 was the largest of all the specimens and specimen 019 was the shortest. After transection of the ACL, the range of anterior tibial translation was 1-5

mm with a median of 3 mm which occurred from 0-30 degrees of flexion. After 30 degrees of flexion, the tibia translated posteriorly for a range of 1-2 mm and the curve parallels that of the intact state. The difference in magnitude between each curve is predominately due to the difference in slopes in the range from 0-20 degrees of flexion. The difference in magnitude between the tibial translation of the intact knees and the ACL deficient knees is largest in the range of 30-70 degrees of flexion with a maximum difference in the range of 30-40 degrees of flexion.

In the range of 0-30 degrees of flexion, tibial translation was relatively unaffected by a change in quadriceps force. In contrast, with the exception of specimen 221, anterior tibial translation of the ACL deficient knees with 40 pounds of quadriceps force were consistently larger than the ACL deficient knees with 10 pounds of quadriceps force. In all cases, between 30 degrees and full flexion, tibial translation of the ACL deficient knees with the 40 pounds quadriceps force closely paralleled the tibial translation of the ACL deficient knees in the normal condition.

C. ROTATION COMPARISONS

Figures 21 through 25 are plots of the relative tibial rotation versus flexion angle of all five specimens. Each curve represents the average rotation of three consecutive half-cycle runs (extension to flexion) at a specific loading condition. Adduction and abduction rotation (rotation about the tibiofemoral direction) was minor compared to the rotation along the tibial axis and will be omitted.

Specimen 005 internally rotated a maximum of 3 degrees relative to the femur in the intact state. In the range of zero to 30 degrees, relative internal tibial rotation gradually increased to its maximum value at 30 degrees of flexion. Beyond 30 degrees, tibial rotation remain constant. At very large flexion angles, internal tibial rotation slightly decreased to a value of 2 degrees. After transection of the ACL, internal tibial rotation slightly decreased below but paralleled the intact state up to 70 degrees. After 70 degrees, internal tibial rotation gradually decreased to a value of 1 degree at full flexion.

Decreasing the quadriceps force to 10 pounds while in the ACL deficient condition significantly increased internal tibial rotation to a value of 6 degrees at 40 degrees of flexion. After 40 degrees, tibial rotation remained a constant 6 degrees to full flexion.

Specimen 066 uncharacteristically rotated externally to a maximum value of 5 degrees at 25 degrees of flexion and remained constant up to 60 degrees of flexion. After 60 degrees, the tibia gradually rotated internally up to 1 degree at full flexion. After transection of the ACL, the tibia rotated very little up to 60 degrees of flexion and beyond 60 degrees the tibia gradually rotated externally up to 6 degrees at full flexion. Increasing the quadricep force to 40 pounds while in the ACL deficient condition slightly decreased internal tibial rotation up to 40 degrees of flexion. After 40 degrees, the curves of the two conditions remained approximately 2 degrees apart and paralleled each other up to full flexion. Decreasing the quadriceps force to 10 pounds significantly increased internal tibial rotation up to a value of 2.5 degrees at 70 degrees of flexion. Beyond 70 degrees, internal tibial rotation paralleled the other ACL deficient conditions.

Specimen 019 externally rotated approximately 8 degrees at full flexion. In the range of zero to 30 degrees, the tibia rotated internally reaching a maximum value of 3 degrees at 30 degrees of flexion. After 30 degrees, the tibia quickly rotated externally reaching its maximum value at full flexion. After transection of the ACL, tibial rotation paralleled the intact state in the range of zero to 30 degrees, and beyond 30 degrees quickly rotated externally reaching its maximum value of 10 degrees at full flexion. Increasing the quadriceps force to 40 pounds while in the ACL deficient condition slightly increased internal tibial rotation throughout the range of motion while decreasing the quadriceps force to 10 pounds dramatically increased internal tibial rotation up to a maximum value of 5 degrees at 30 degrees of flexion up to full flexion.

At very large flexion angles, this condition became very unstable and tibial rotation behaved unpredictably. Specimen 037 externally rotated approximately 10 degrees at full flexion. In the range of zero to 80 degrees, internal tibial rotation reached approximately 1 degree and changed very little. After 80 degrees, the tibia quickly rotated

D. COUPLED TRANSLATION AND ROTATION

It is well documented that the basic mechanism of movement between the femur and the tibia is a combination of rolling and gliding. Presently, there are no models that can predict the exact mix of rolling and gliding or when they occur throughout the range of motion. The difficulty arises because automatic initial and terminal rotation as well as voluntary rotation are superimposed throughout the range of motion in the sagittal plane; the plane in which the majority of knee motion is occurring. However, the six degree of freedom model does not restrict our analysis to knee motion in the sagittal plane. Thus, by synchronizing the translation plots with the rotation plots, an estimate can be made as to the mix of rolling and gliding as well as to when each occurs during flexion.

Figures 26 through 28 are plots showing the average anterior tibial translation and relative tibial rotation versus flexion angle. The intact and ACL deficient knee in the normal condition was chosen to simplify the analysis. Additionally, the translation plots of specimens 037, 221, and 019 exhibited similar behavior and make qualitative comparisons easier.

In the range of 0-20 degrees of flexion, the three specimens translate with a negative gradient, i.e., the tibia translate posteriorly relative to the femur. The range of posterior tibial translation is approximately 2.5 mm in specimens 037 and 221 and 7.5 mm in specimen 019. In the same range of flexion, the three specimens underwent relative rotation with a negative gradient. The range of relative rotation is approximately 1-2 degrees for specimens 037 and 221, and 5 degrees for specimen 019. In the range of 80 degrees to full flexion, the specimens translate with a large positive gradient reaching a maximum translation at full flexion. The relative rotation rapidly climbs to their maximum values.

This type of motion is characteristic of the screw-home mechanism; a combination of knee extension and external tibial rotation or unscrew for knee flexion and internal tibial rotation. In both cases, the shape of the femoral condyles as well as the asymmetry of the tibiofemoral joint serve to increase the stability of the knee at full

extension by locking the articular surfaces of the joint. Contrastingly, the same mechanism in reverse places the anterior cruciate ligament in a vulnerable position during the early phases of flexion, where the ACL is most effective in limiting anterior tibial translation. The spiral motion of the tibia about the femur during the locking and unlocking phases of extension and flexion is a direct result of the medial femoral condyle being larger than the lateral femoral condyle. As the tibia translates relative to the femur from the fully extended to the fully flexed position, it ascends and then descends the curves of the medial femoral condyle and simultaneously rotates internally. This motion is reversed as the tibia moves back into the fully extended position. The internal rotation of the tibia due to the unlocking of the tibiofemoral joint contributes to the majority of the posterior tibial translation observed in the range of 0-15 degrees of flexion (recall that the reference point for tibial translation is located on the medial side of the femoral axis). The extent to which the tibia rotates internally during this phase of motion depends on the effectiveness of the ACL and lateral collateral ligaments in limiting this motion, and to some extent, the degree of varus or valgus of the donor's knees. The remaining posterior tibial translation can be attributed to rolling of the tibia to ascend the medial femoral condyle. Very little glide can occur in this region of flexion since the knee must unlock itself prior to gliding. The mechanism of gliding is evident in the translation plots where tibial translation is relatively constant.

Once the ACL was transected, the tibia continued to roll on the medial femoral condyle for an additional 5-8 degrees of flexion before it began to glide. At this point, tibial translation of the ACL transected knee remained posterior to the tibial translation of the intact knee between 75-80 degrees of flexion. Beyond 80 degrees of flexion, the difference between the translations decreased. This effect may be attributed to the fact that the range of flexion in which the ACL is most effective in limiting anterior tibial translation is 0-40 degrees. Hence, ACL insufficiency results in a disintegration of the rolling-gliding movement, causing the femur to roll excessively on the tibia before it glides.

In the range of 80-90 degrees of flexion, the movement of the tibia is a mixture of gliding and rolling, with the ratio of rolling to gliding being much less than the ratio in the early phases of flexion. The contribution to tibial translation from relative rotation of the tibia is less evident in this phase of flexion, but must be assumed present. For the sake of completeness, other factors such as the shape of the femoral condyles, tibial plateau, menisci, and cruciate ligament insertion sites must be addressed using empirical methods.

In summary, the six degree of freedom model qualitatively predicted the range of flexion in which the ACL is most effective in limiting anterior tibial translation as well as the pathokinematics of an ACL deficient knee. Moreover, the model enhanced the capability to "see" the disintegration of the rolling-gliding movement in an ACL deficient knee throughout the range of motion. The ACL functions as the primary restraint to limit anterior tibial translation while offering no restraint to posterior tibial translation. Sectioning of the ACL allowed the femur to roll excessively on the tibia before gliding while increasing internal tibial rotation during flexion. Hence, the ACL functions as a secondary restraint to limit internal tibial rotation while offering no restraint to external rotation.

E. ESTIMATED MOTION OF INSTANT CENTER

The instant center of the tibiofemoral joint provides a description of the relative planar motion of two adjacent segments of a body and the direction of displacement of the contact points between these segments. O'Conner [Ref. 10] defines the instant center of the joint as that point at which the neutral fibers of the ACL and PCL cross. The perpendicular line passing through this point is defined as the flexion axis. Thus, the instant center is the geometric center of the knee in the sagittal plane. In principle, each knee could then be identified by the motion of its own unique instant center. O'Conner further describes that the flexion axis moves relative to both bones during flexion or extension, i.e., the path that the instant center follows relative to the femur is different than the path followed relative to the tibia. This phenomenon can be validated by fixing

the tibia or femur and allowing the other bone to move. Although this six degree of freedom model can not predict the location of the instant center, the data obtained using the translation transducers, which are fixed to the medial penetration point of the femoral transepicondylar line, was used as a rough estimate of the motion of the instant center. This was based on the assumption that the orientation of the femoral link to the tibial link remains relatively constant during flexion and that motion in the transverse and frontal planes is small compared to the motion in the sagittal plane. The data recorded, in relation to the transverse and frontal planes, was small in magnitude than those recorded in the sagittal plane. Once the motion of the instant center of an intact knee is known, it can be used determine the effect of cruciate injury and effectiveness of surgical reconstruction techniques.

Figures 29 through 33 are plots of the estimated motion of the instant center in an intact and ACL deficient knee. The normal condition of each specimen was used for the analysis. The estimated motion of the instant center along the medial-lateral direction consist of a component of tibial rotation as well as a component of adduction-abduction rotation. The remaining component of tibial rotation contributes to tibial translation. The plots show that the component of tibial rotation that contributes to anterior-posterior displacement is larger than the component of tibial rotation that contributes to medial-lateral displacement. Because of the additional restraint imposed on knee motion by the medial and lateral collateral ligaments, adduction-abduction rotation is much smaller than tibial rotation. This restraint would make the contribution to medial-lateral displacement by the component of tibial rotation along the medial-lateral direction more significant than the contribution by adduction-abduction rotation. The plots of the motion of the instant center in the transverse plane show that four of the five specimens experienced a significant lateral shift upon transection of the ACL (minimum shift in specimen 037). The lateral shift occurs early in the phase of flexion of the ACL deficient knee and then abruptly changes direction toward the motion of the intact state. After this abrupt change in direction, the femur and tibia are again in normal relation to each other for the degree

of flexion attained. The lateral shift demonstrates the disintegration of the rolling-gliding movement as a consequence of ACL insufficiency.

F. TOTAL KNEE REPLACEMENT

Upon completion of the ACL deficient tests, the knees were totally reconstructed using two methods of total joint arthroplasty. The first method inserted the prosthesis around the intact PCL. This technique is identified as "PCL retaining" total knee replacement (TKR). The second method sacrificed the PCL before the prosthesis was inserted and is identified as "PCL sacrificing" TKR. Upon completion of the total knee replacements, the kinematics of the specimens were recorded and compared to the kinematics of the corresponding intact state. Complete data sets were recorded for four knee specimens. Figure 34 is a sample plot of the total knee replacement of specimens 005 and 221. Neither TKR technique restored the kinematics to the intact state. The femurs of the TKRs rolled further than the femurs of the intact state in the range of motion in which the ACL is most effective in limiting anterior tibial translation. Beyond 30 degrees of flexion, the tibias remained posteriorly displaced relative to the femur. Beyond 40 degrees of flexion, the PCL retaining TKR technique paralleled the intact state while the PCL sacrificing TKR technique rolled further before paralleling the other two states. The disintegration of the rolling-gliding movement seems to be exacerbated by the total knee replacements while recovering very little of the kinematics of the intact state. The PCL retaining TKR resulted in knee motion that more closely approximated the intact knee.

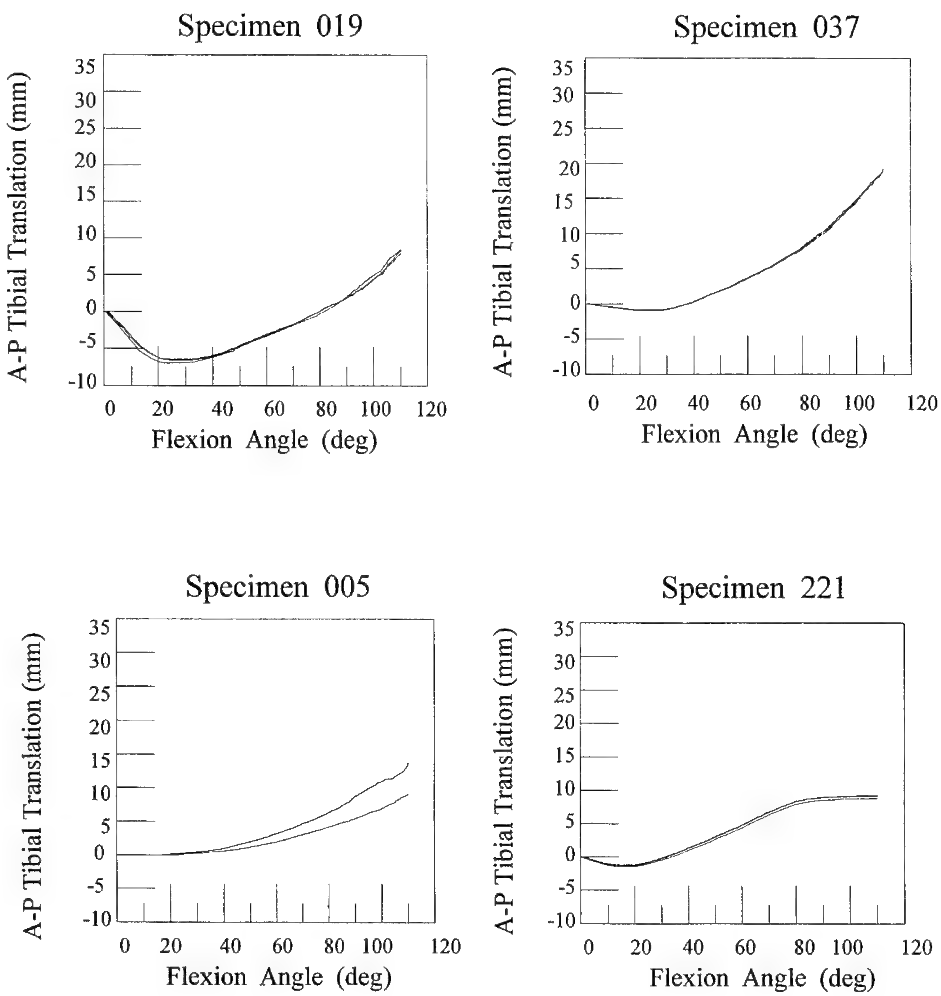

Figure 7: Tibial Translation of Specimens 019, 037,005, 221 During Three Consecutive Half-Cycle Runs (Extension to Flexion) in the Normal Condition.

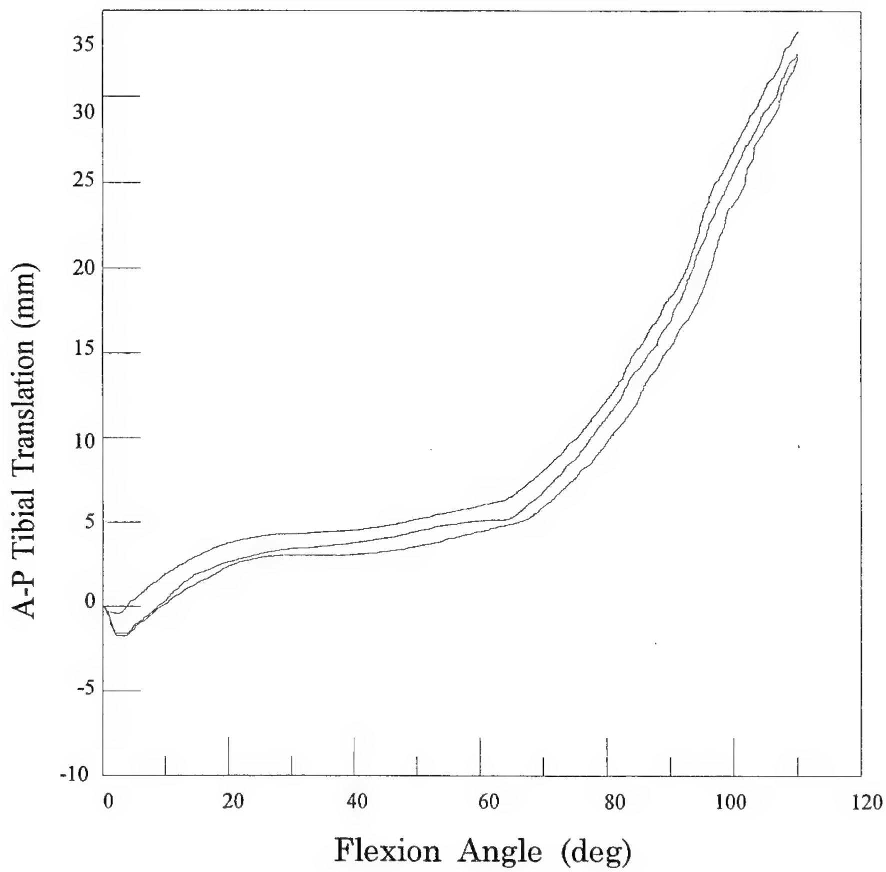

Figure 8: Tibial Translation of Specimen 066 During Three Consecutive Half-Cycle Runs (Extension to Flexion) in the Normal Condition.

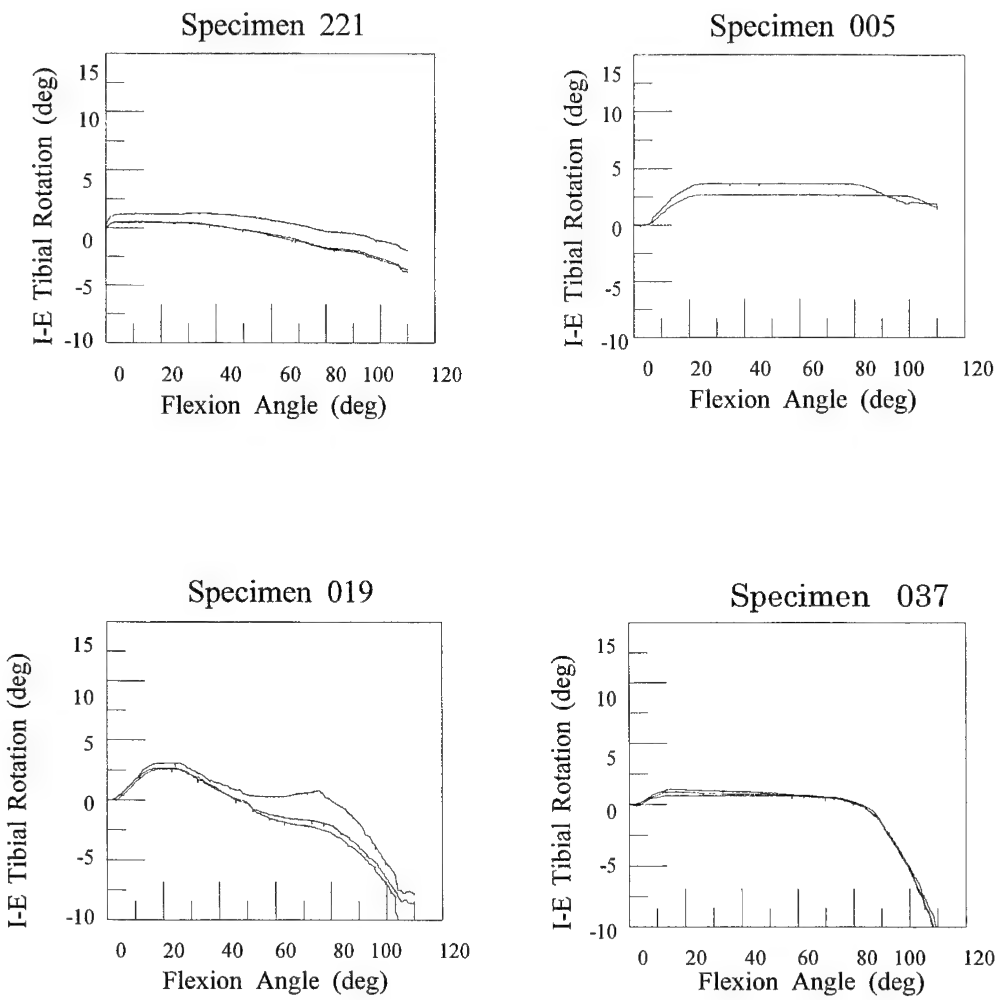

Figure 9: Tibial Rotation of Specimens 005, 019, 037, and 221 During Three Consecutive Half-Cycle Runs (Extension to Flexion) in the Normal Condition.

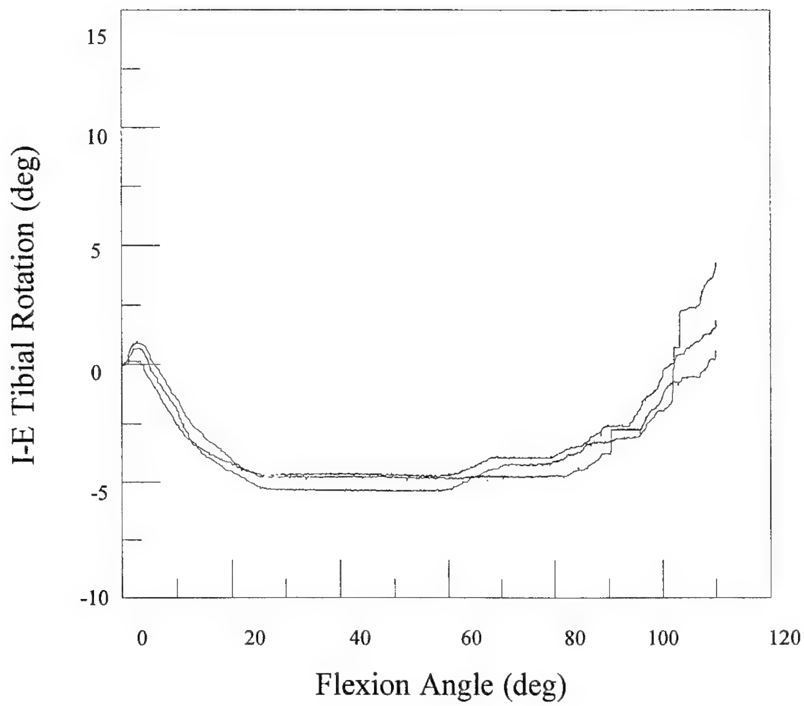

Figure 10: Tibial Rotation of Specimen 066 During Three Consecutive Half-Cycle Runs (Extension to Flexion) in the Normal Condition.

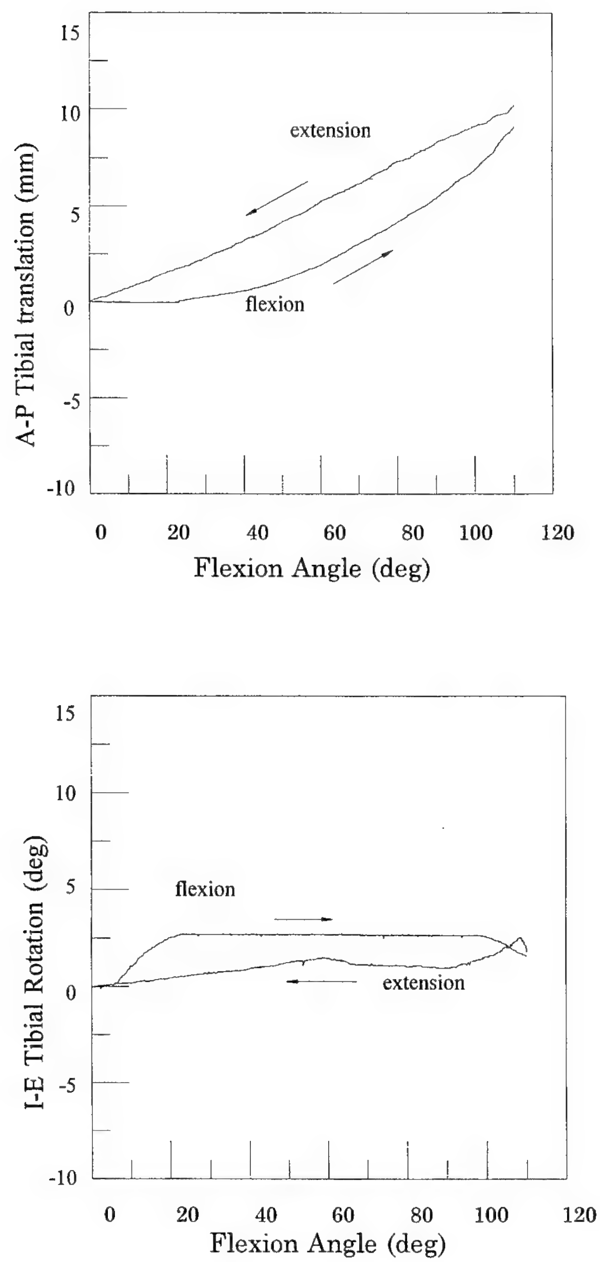

Figure 11: Hysteresis Loop of Specimen 005 in the Normal Condition.

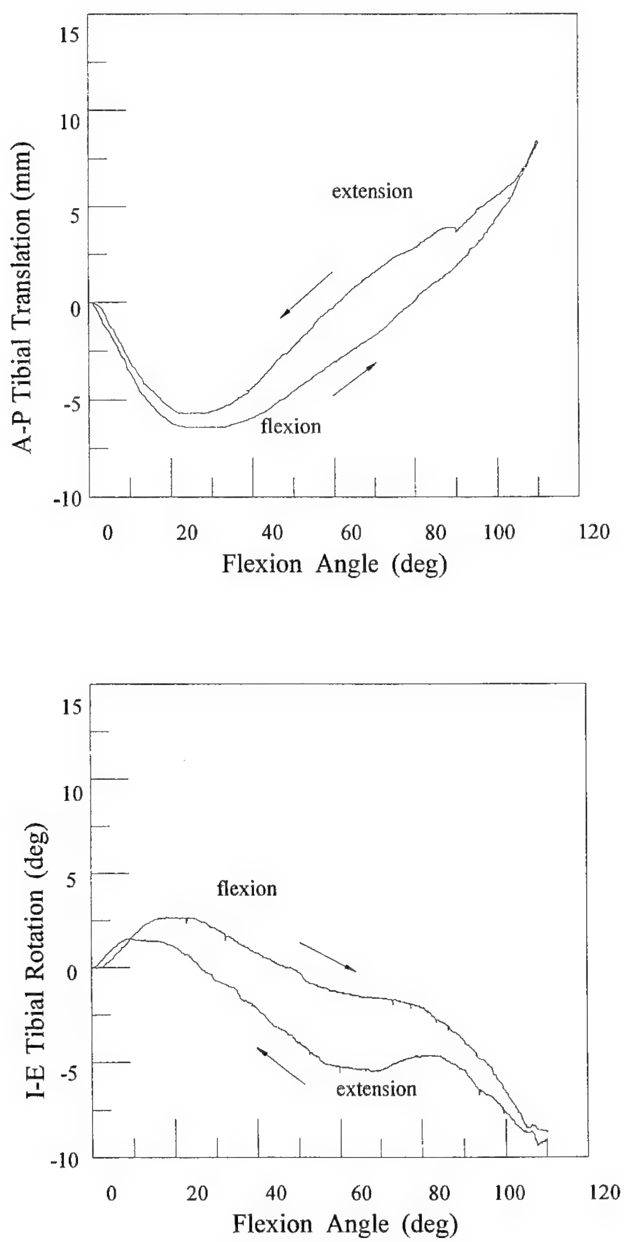

Figure 12: Hysteresis Loop of Specimen 019 in the Normal Condition.

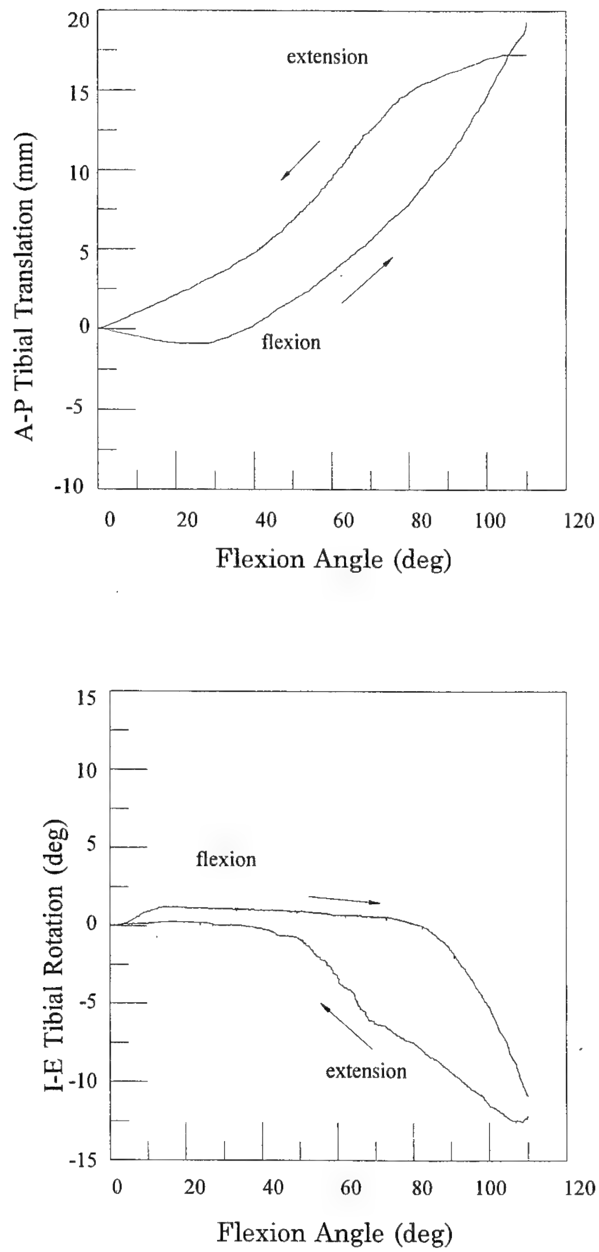

Figure 13: Hysteresis Loop of Specimen 037 in the Normal Condition.

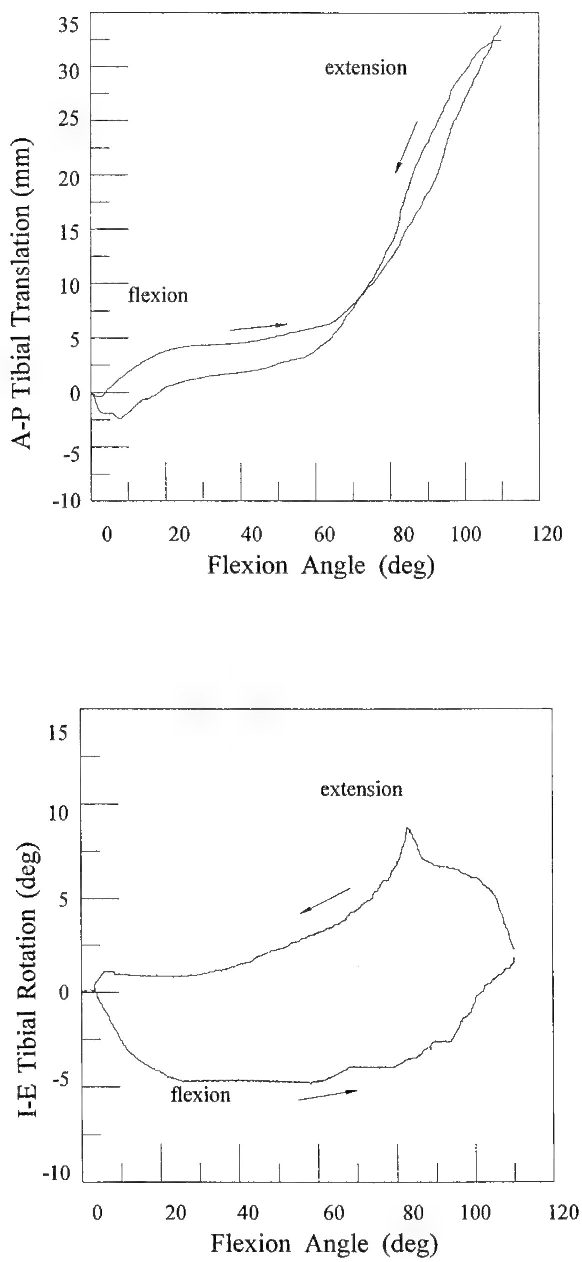

Figure 14: Hysteresis Loop of Specimen 066 in the Normal Condition.

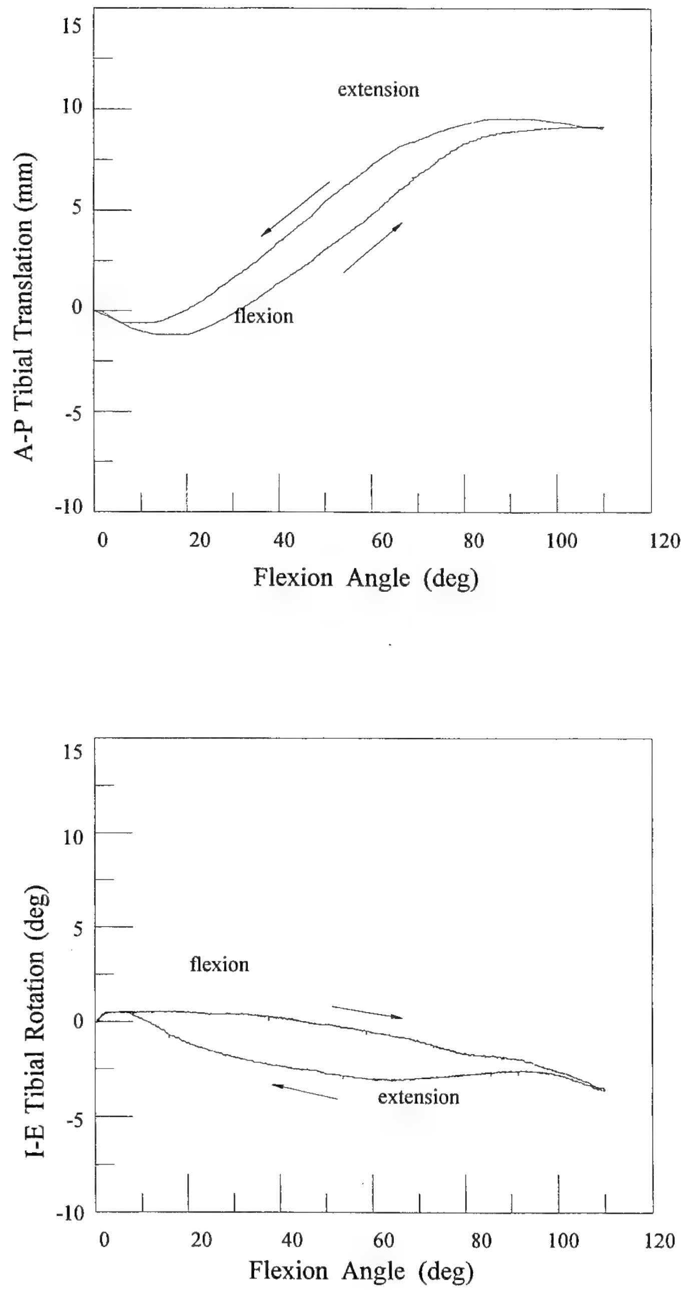

Figure 15: Hysteresis Loop of Specimen 221 in the Normal Condition.

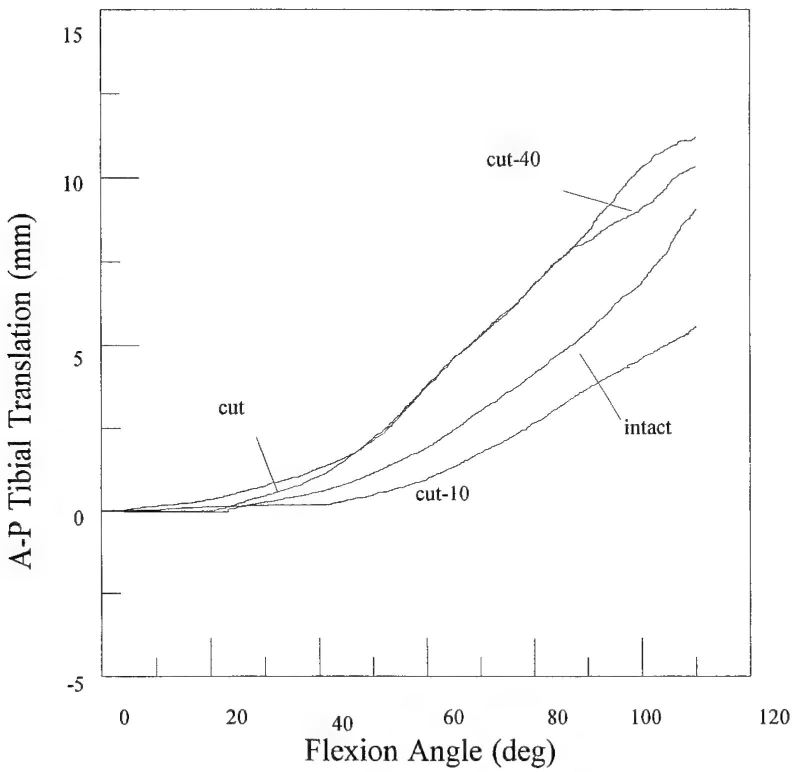

Figure 16: Tibial Translation of Specimen 005 in Flexion. The Plot Shows the Specimen in the Intact, ACL Deficient, ACL Deficient with a 40 Pounds Quadriceps Force, and ACL Deficient with a 10 Pound Quadriceps Force Condition.

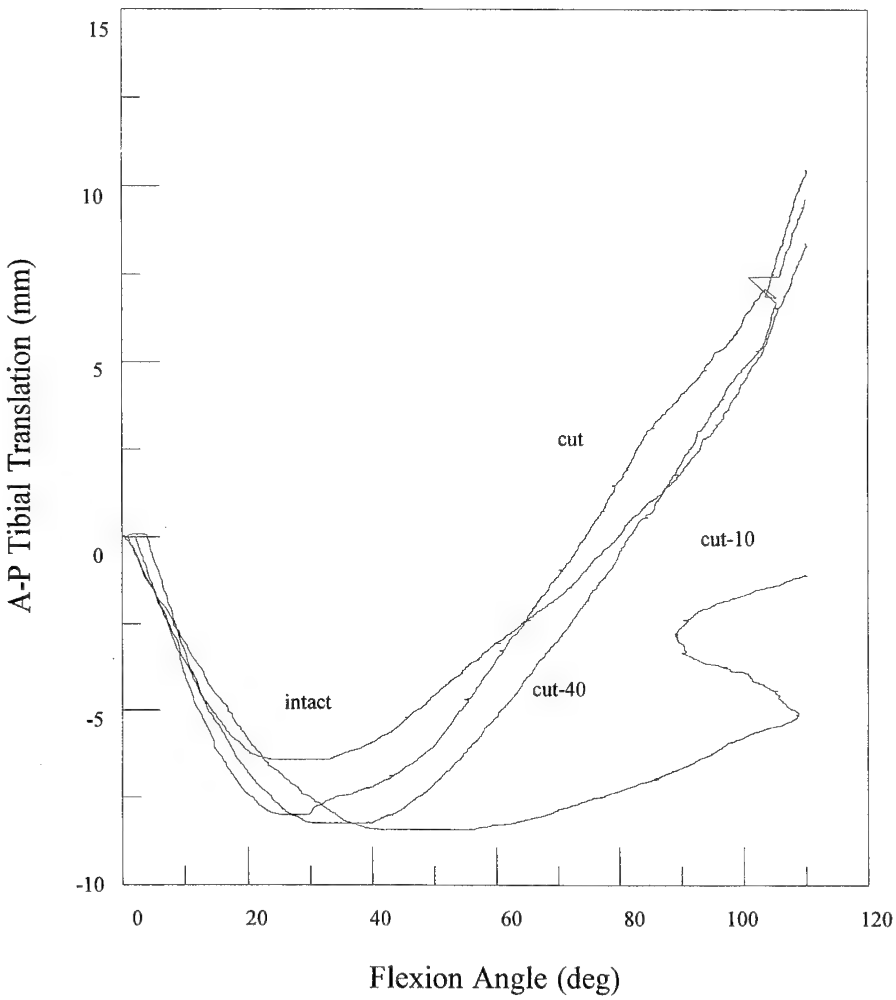

Figure 17: Tibial Translation of Specimen 019 in Flexion. The Plot Shows the Specimen in the Intact, ACL Deficient, ACL Deficient with a 40 Pounds Quadriceps Force, and ACL Deficient with a 10 Pound Quadriceps Force Condition.

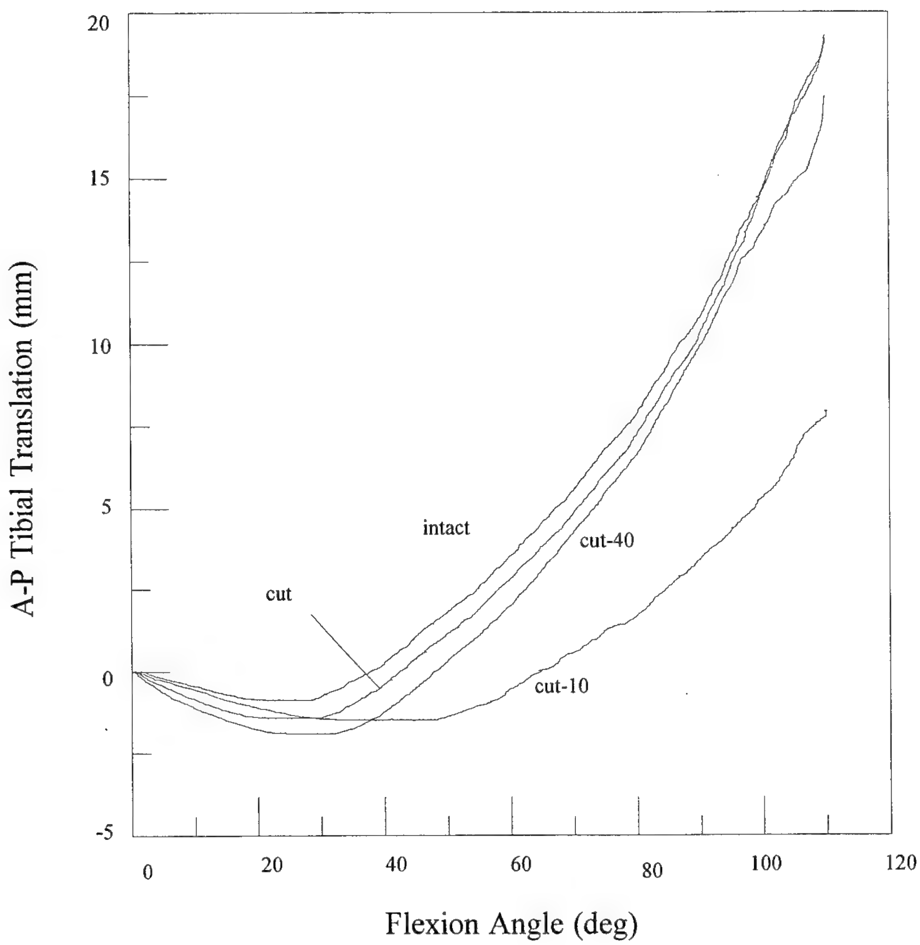

Figure 18: Tibial Translation of Specimen 037 in Flexion. The Plot Shows the Specimen in the Intact, ACL Deficient, ACL Deficient with a 40 Pound Quadriceps Force, and ACL Deficient with a 10 Pound Quadriceps Force Condition.

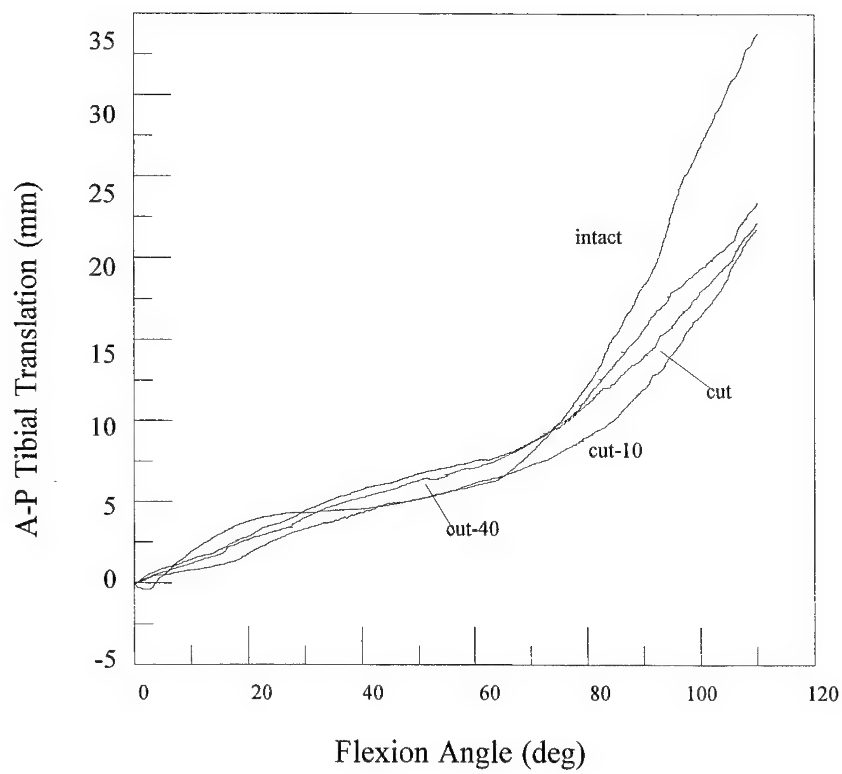

Figure 19: Tibial Translation of Specimen 066 in Flexion. The Plot Shows the Specimen in the Intact, ACL Deficient, ACL Deficient with a 40 Pound Quadriceps Force, and ACL Deficient with a 10 Pound Quadriceps Force Condition.

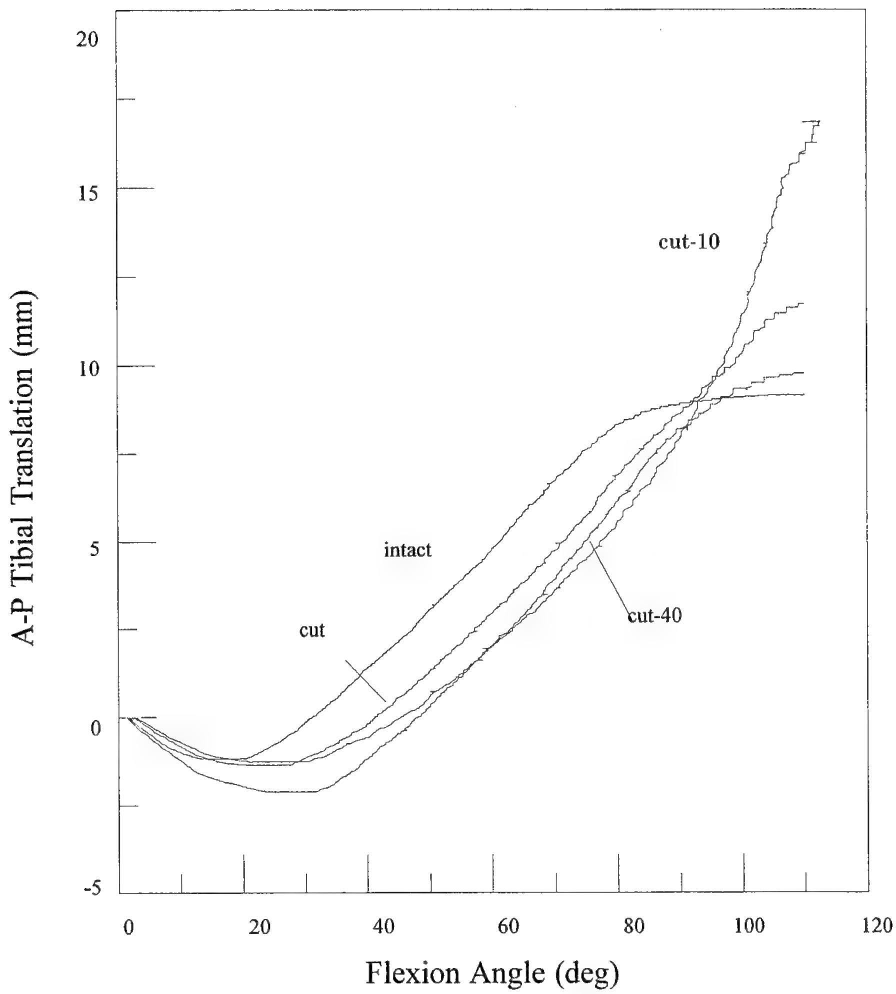

Figure 20: Tibial Translation of Specimen 221 in Flexion. The Plot Shows the Specimen in the Intact, ACL Deficient, ACL Deficient with a 40 Pound Quadriceps Force, and ACL Deficient with a 10 Pound Quadriceps Force Condition.

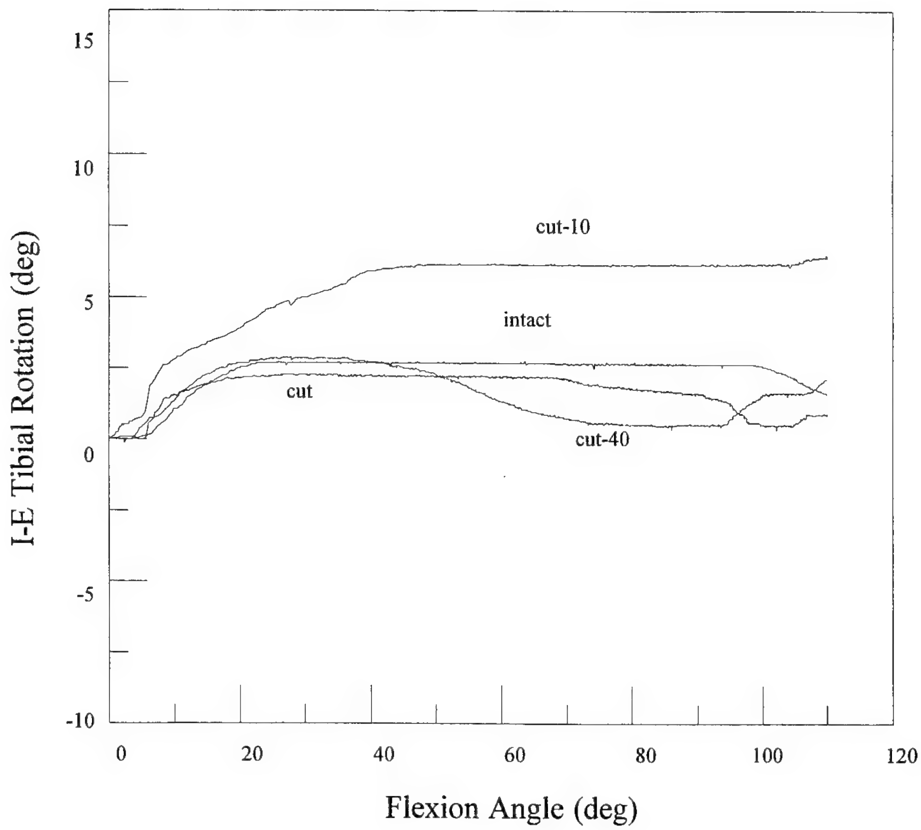

Figure 21: Tibial Rotation of Specimen 005 in Flexion. The Plot Shows the Specimen in the Intact, ACL Deficient, ACL Deficient with a 40 Pound Quadriceps Force, and ACL Deficient with a 10 Pound Quadriceps Force Condition.

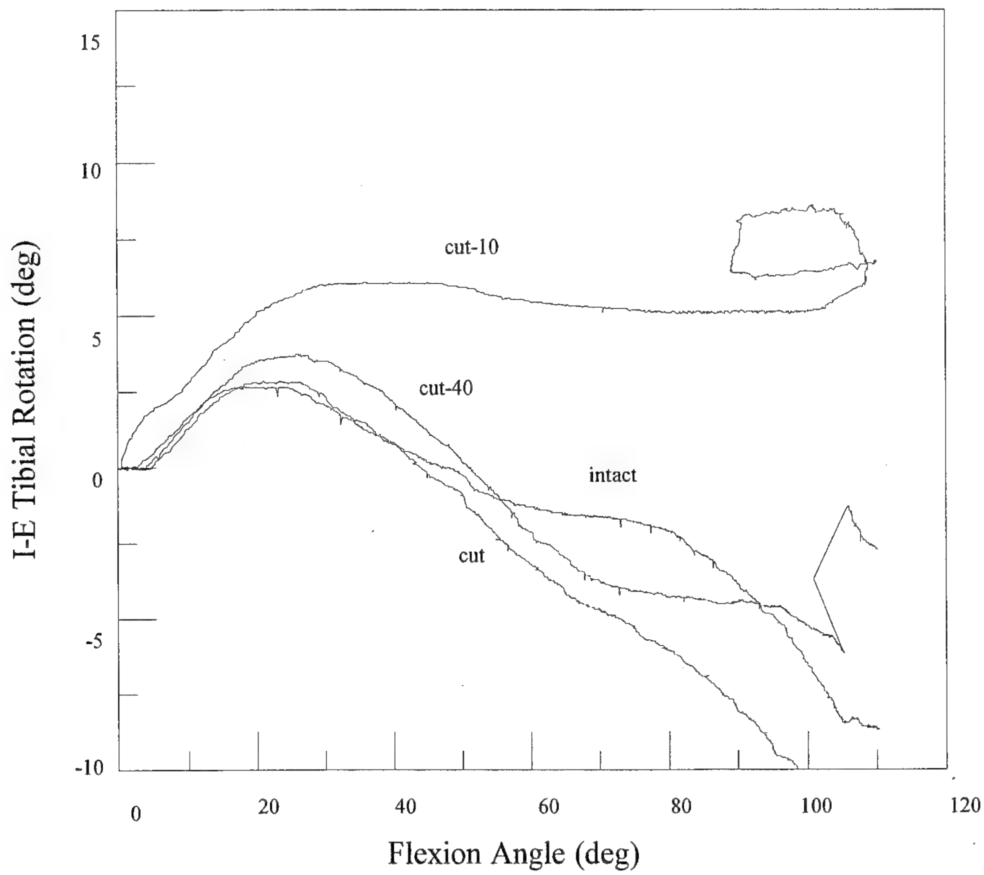

Figure 22: Tibial Rotation of Specimen 019 in Flexion. The Plot Shows the Specimen in the Intact, ACL Deficient, ACL Deficient with a 40 Pound Quadriceps Force, and ACL Deficient with a 10 Pound Quadriceps Force Condition.

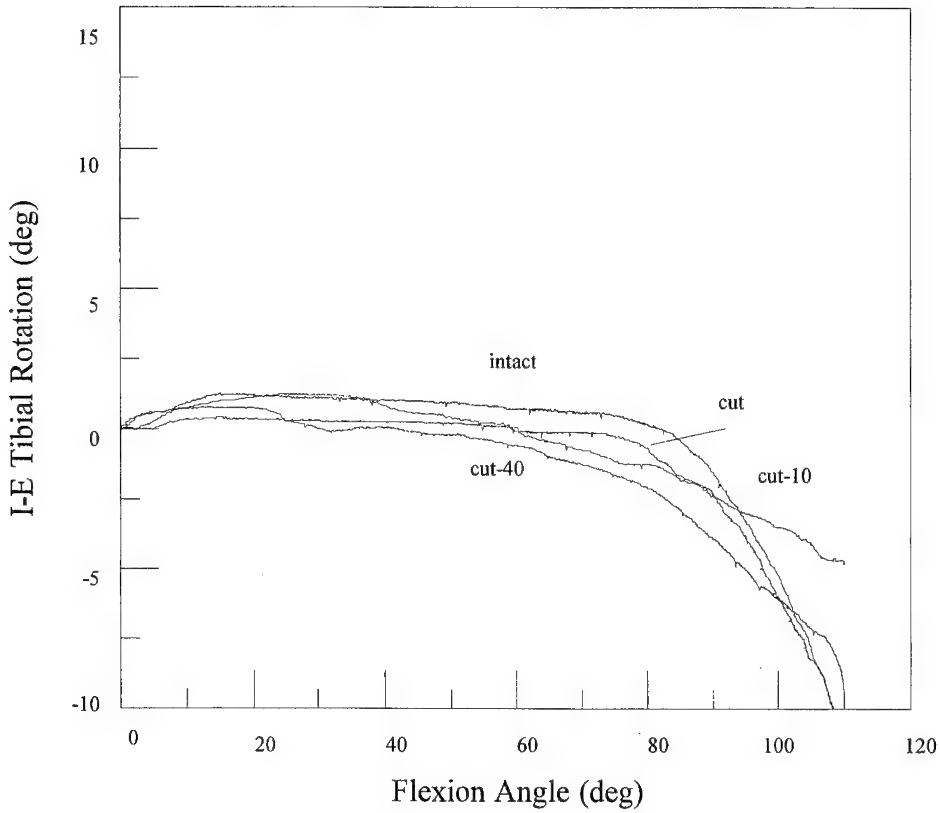

Figure 23: Tibial Rotation of Specimen 037 in Flexion. The Plot Shows the Specimen in the Intact, ACL Deficient, ACL Deficient with a 40 Pound Quadriceps Force, and ACL Deficient with a 10 Pound Quadriceps Force Condition.

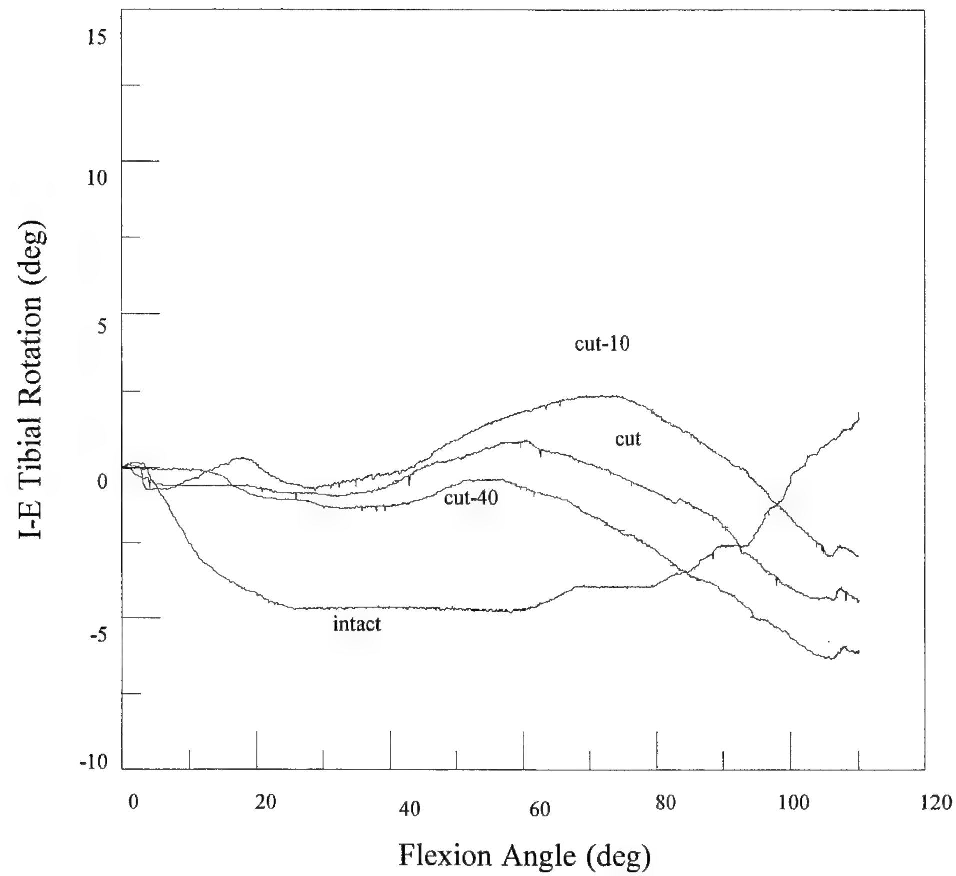

Figure 24: Tibial Rotation of Specimen 066 in Flexion. The Plot Shows the Specimen in the Intact, ACL Deficient, ACL Deficient with a 40 Pound Quadriceps Force, and ACL Deficient with a 10 Pound Quadriceps Force Condition.

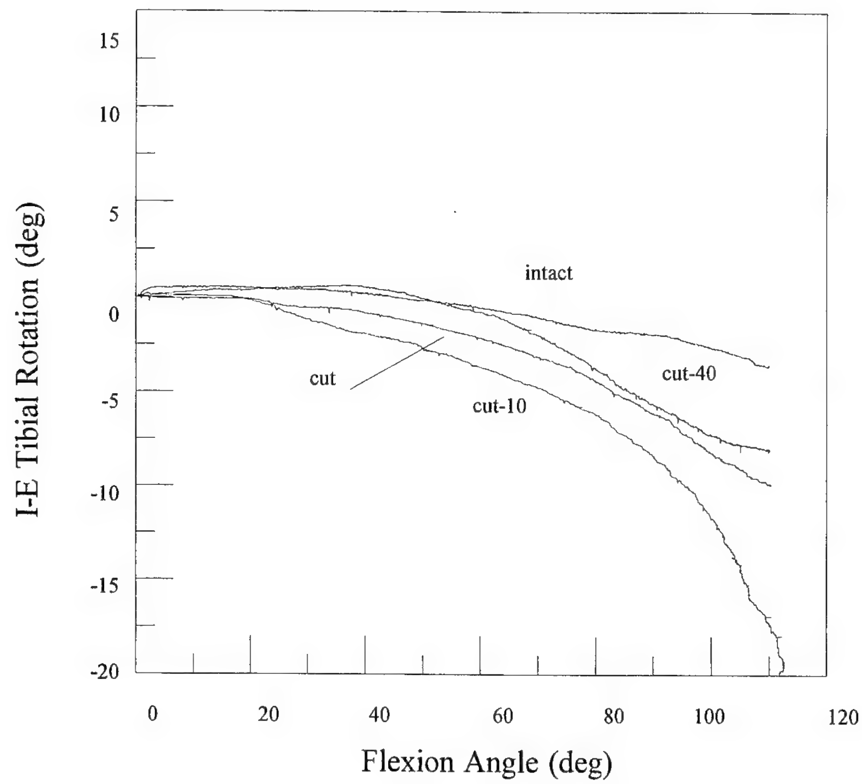

Figure 25: Tibial Rotation of Specimen 221 in Flexion. The Plot Shows the Specimen in the Intact, ACL Deficient, ACL Deficient with a 40 Pound Quadriceps Force, and ACL Deficient with a 10 Pound Quadriceps Force Condition.

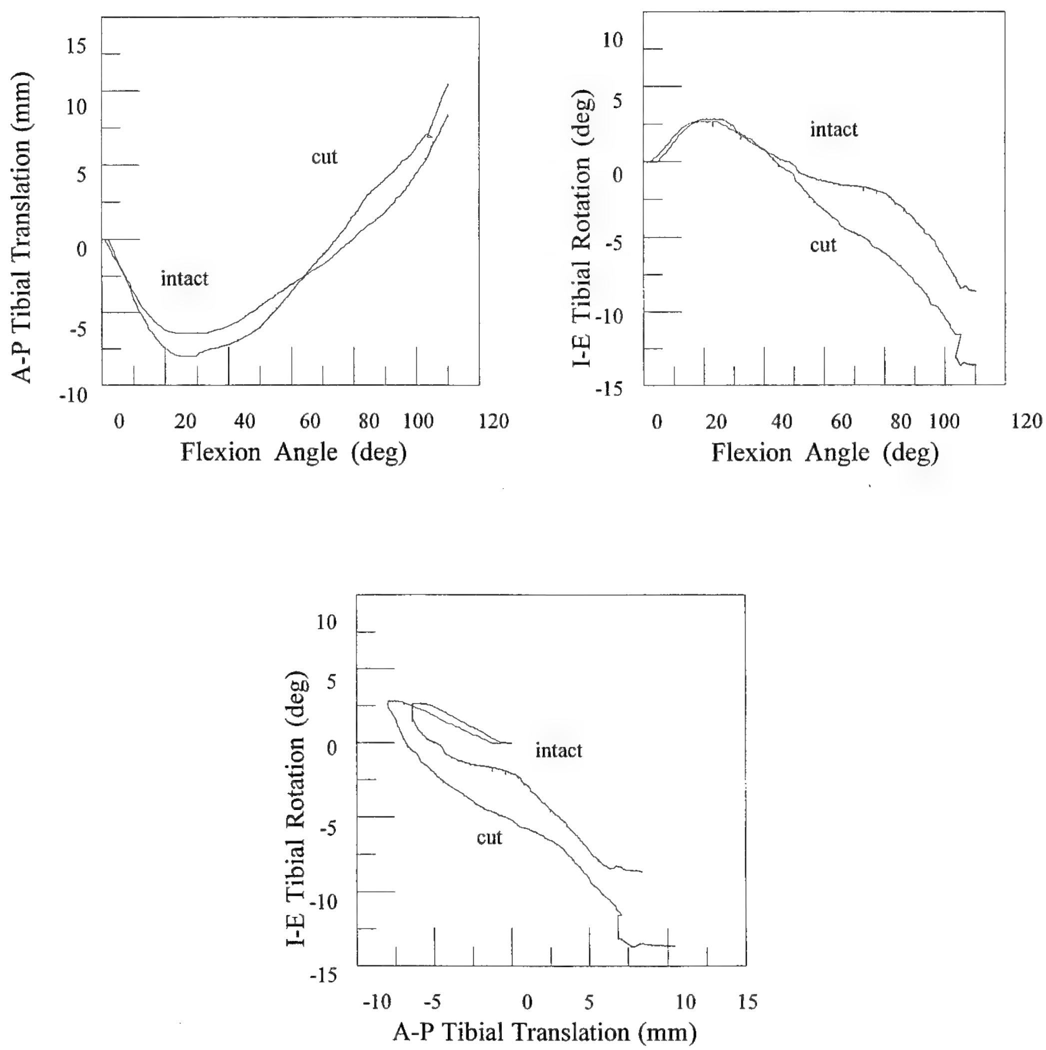

Figure 26: Intact Versus ACL Deficient State of Specimen 019 in the Normal Condition.

The Plots Show the Influence of Tibial Rotation on Translation Along the Anterior-Posterior Drawer.

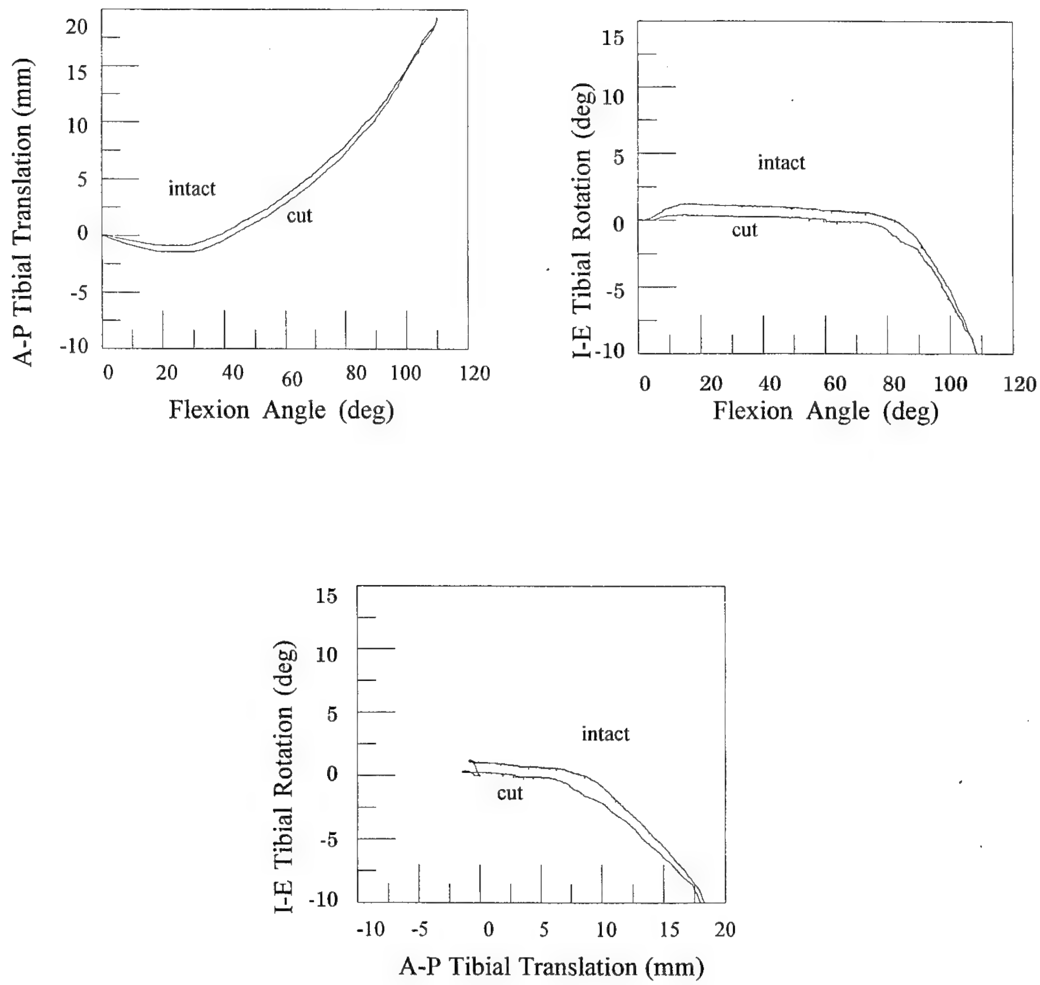

Figure 27: Intact Versus ACL Deficient State of Specimen 037 in the Normal Condition. The Plots Show Influence of Tibial Rotation on Translation Along the Anterior-Posterior Drawer.

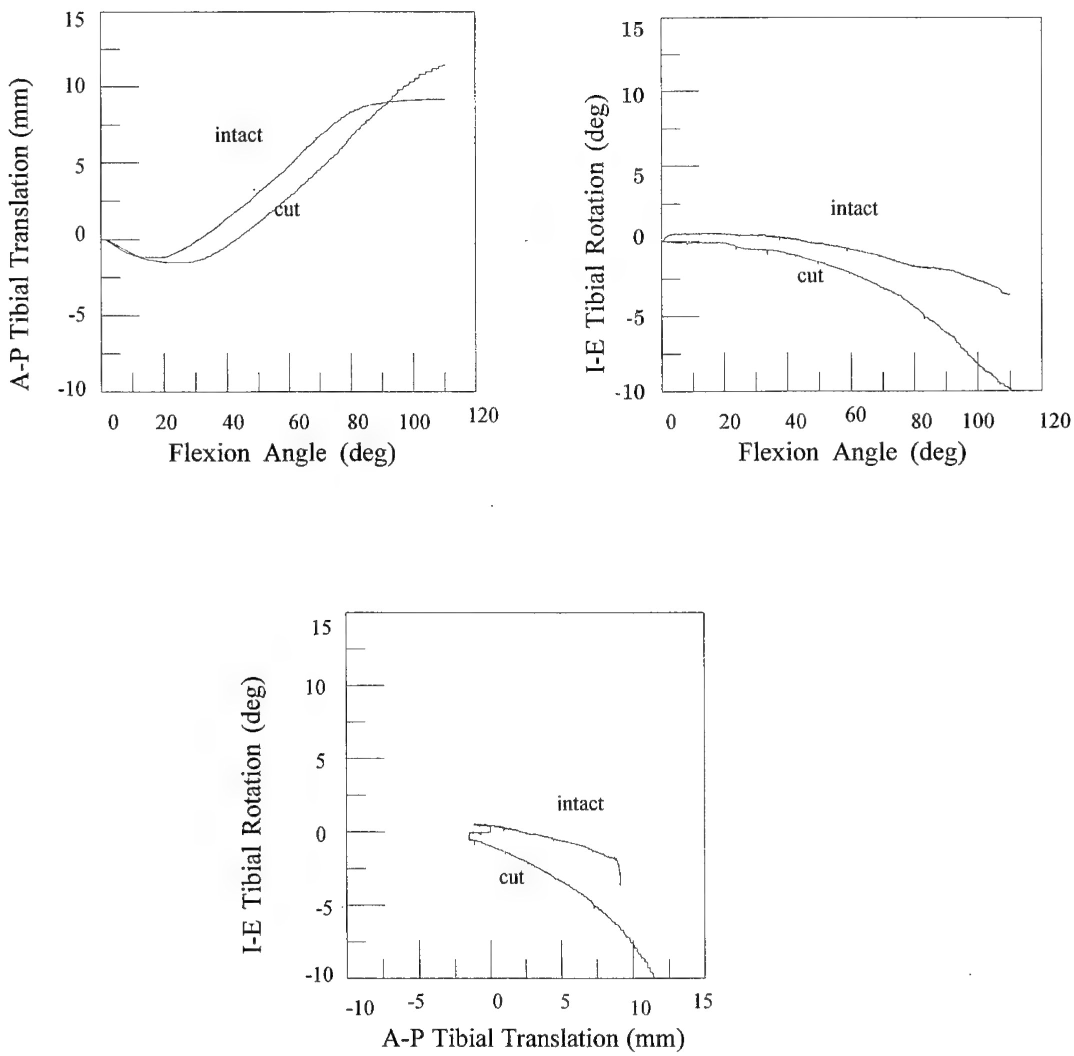

Figure 28: Intact Versus ACL Deficient State of Specimen 221 in the Normal Condition. The Plots Show the Influence of Tibial Rotation on Translation Along the Anterior-Posterior Drawer.

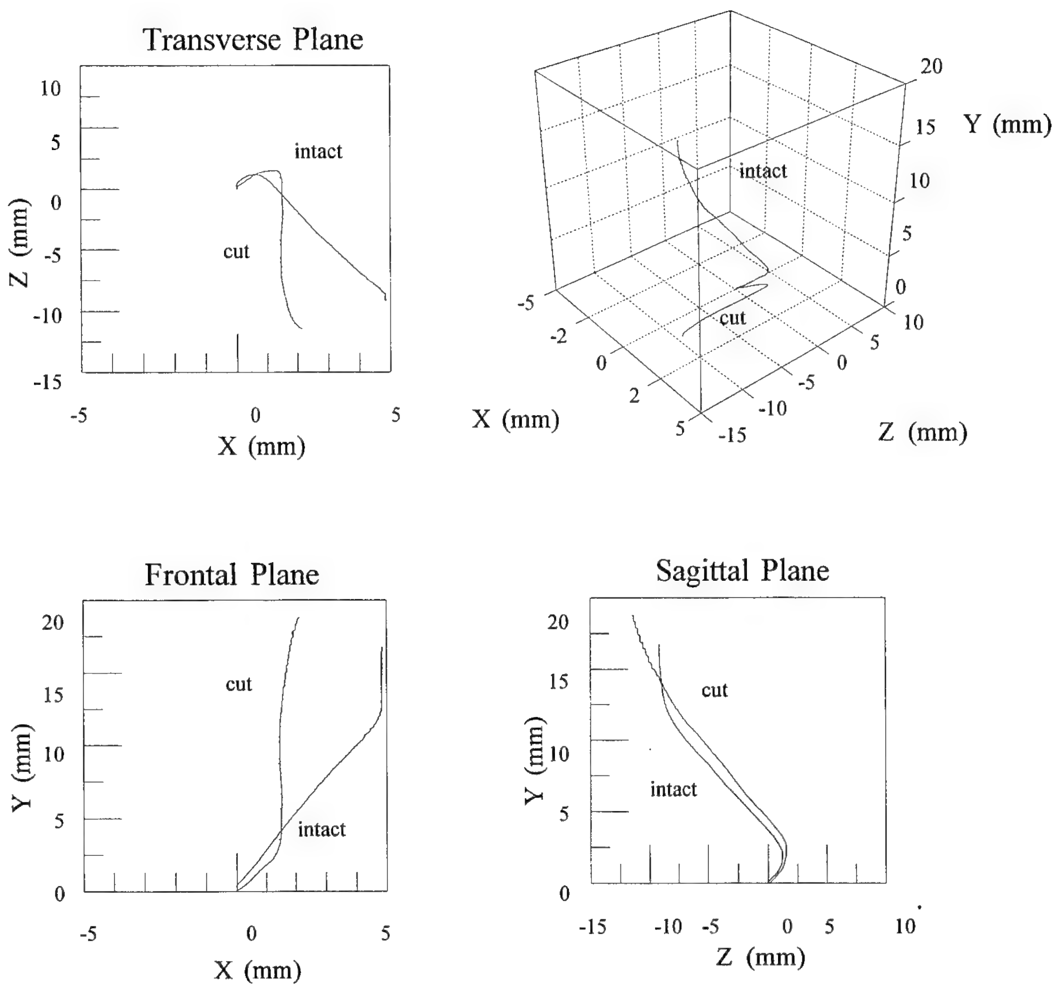

Figure 29: Comparison of the Intact and ACL Deficient State of the Estimated Motion of the Instant Center of Specimen 221 in the Normal Condition.

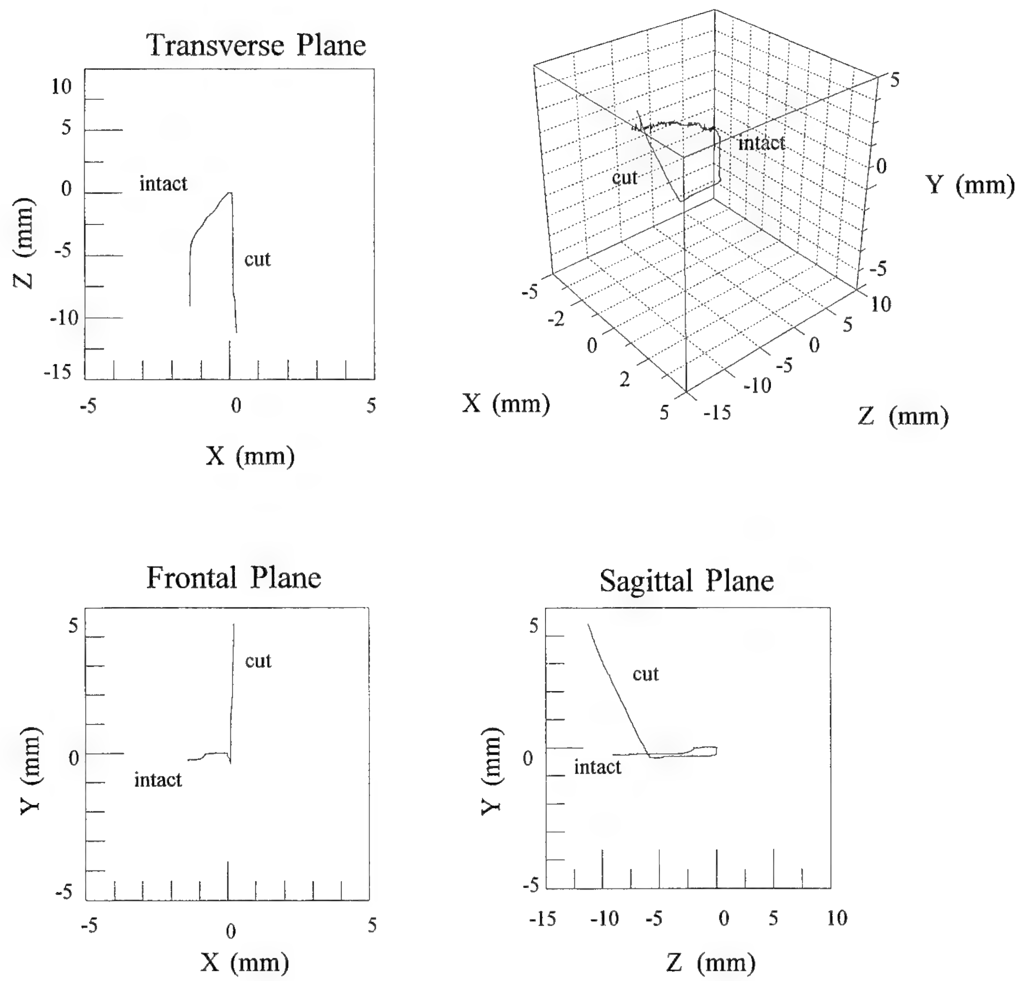

Figure 30: Comparison of the Intact and ACL Deficient State of the Estimated Motion of the Instant Center of Specimen 005 in the Normal Condition.

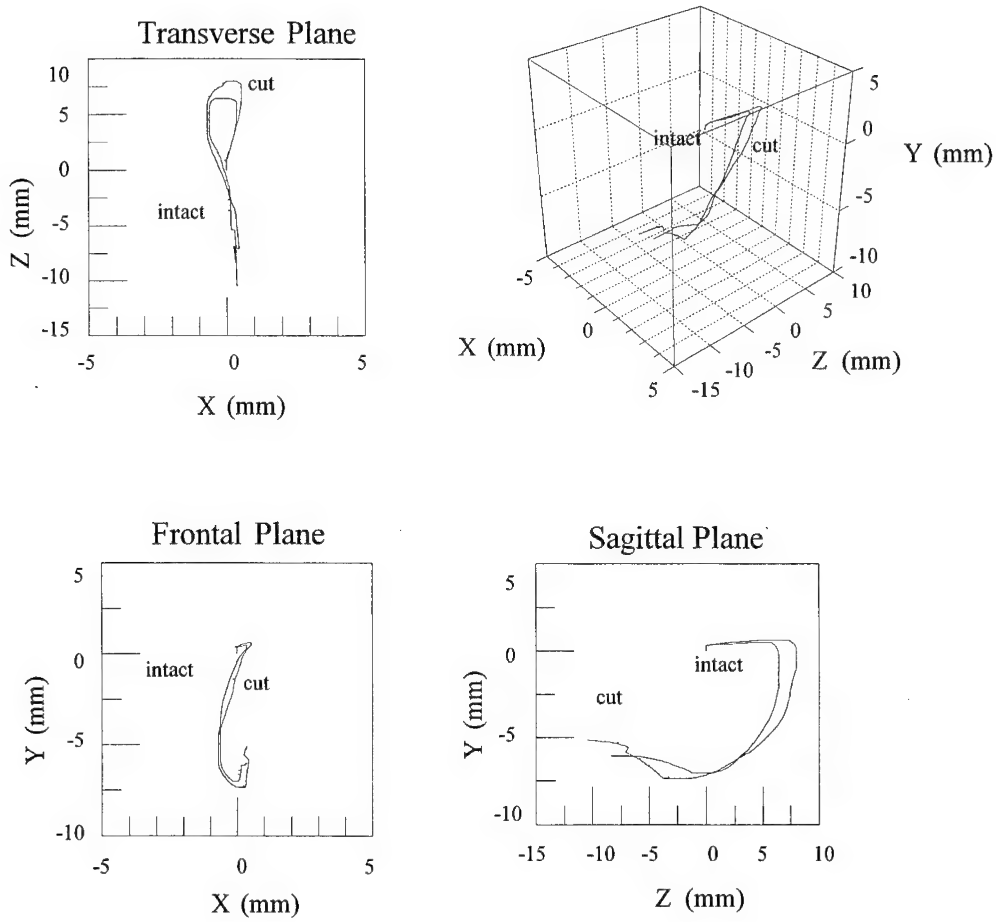

Figure 31: Comparison of the Intact and ACL Deficient State of the Estimated Motion of the Instant Center of Specimen 019 in the Normal Condition.

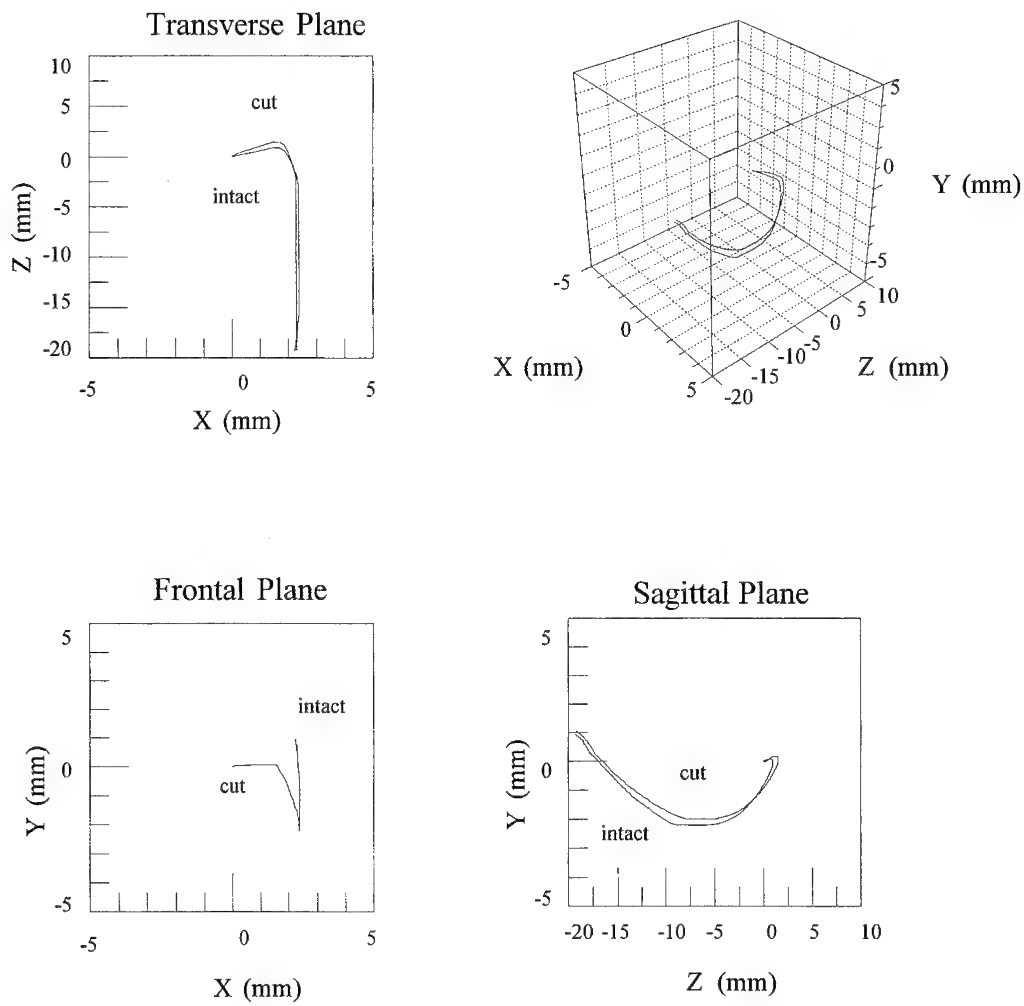

Figure 32: Comparison of the Intact and ACL Deficient State of the Estimated Motion of the Instant Center of Specimen 037 in the Normal Condition.

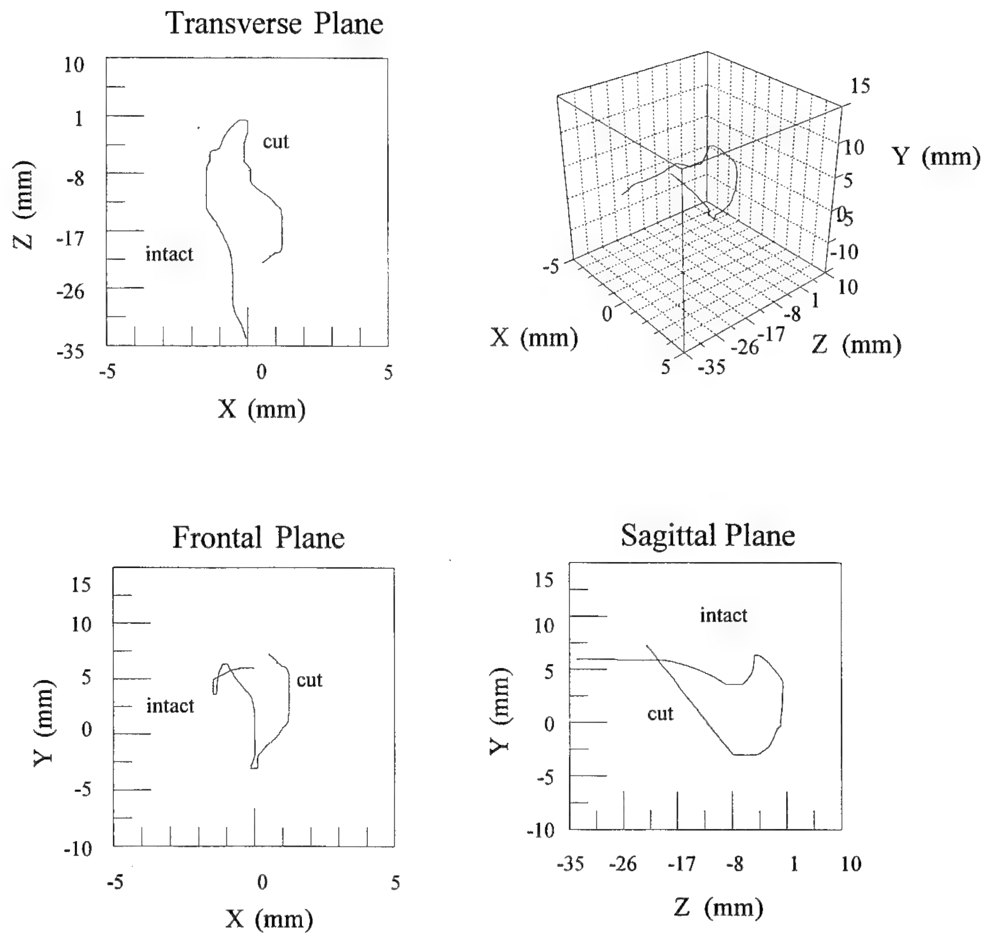

Figure 33: Comparison of the Intact and ACL Deficient State of the Estimated Motion of the Instant Center of Specimen 066 in the Normal Condition.

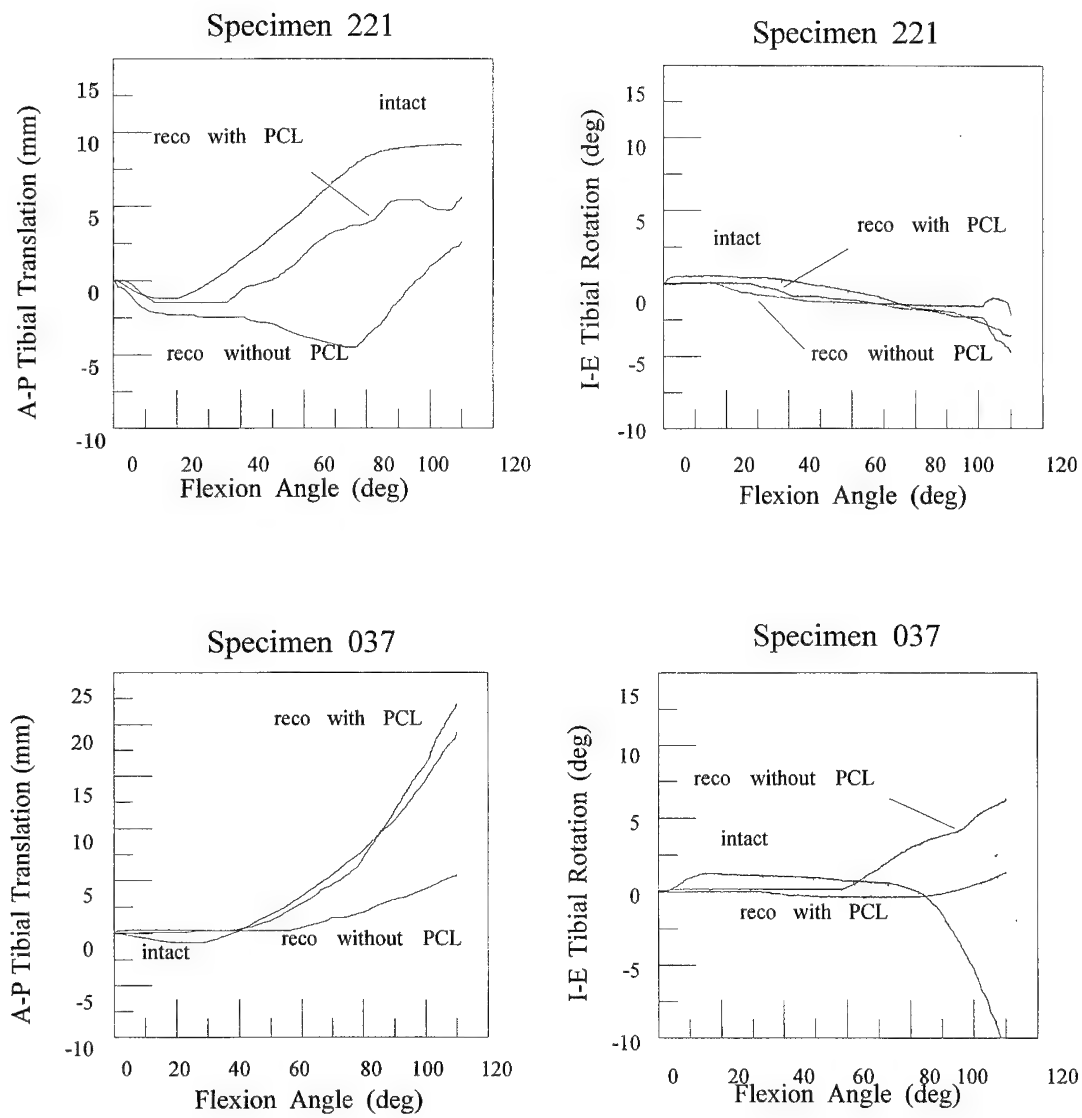

Figure 34: Comparison of Total Knee Replacement Techniques with the Corresponding Intact State. The Plots Reflect the Normal Condition.

V. CONCLUSIONS

This testing method allowed continuous motion, continuous loading, unconstrained data acquisition, and six degrees of freedom to reliably measure the kinematics of the intact and cruciate deficient cadaveric knee. From the data it was shown that the ACL functions primarily as a restraint to limit anterior tibial translation. Once the ACL was transected, the range of anterior tibial translation increased in the range of 0-30 degrees of flexion. Past 30 degrees, the tibia translates anteriorly relative to the femur and paralleled the intact state. Moreover, the data demonstrated that a secondary role of the ACL is to act as a constraint to limit internal tibial rotation during the early phase of flexion. Once the ACL was transected, the range of internal tibial rotation was larger for the knee in the ACL deficient state than in the intact state. Absence of the ACL resulted in disintegration of the rolling-gliding movement causing the femur to roll excessively on the tibia before it glides.

Estimated motion of the instant center of the tibiofemoral joint can be approximated by observing the motion of a point on the femoral transepicondylar line which acts as the flexion axis in three dimensions. After transection of the ACL, estimated motion of the instant center in the transverse plane, although small compared to motion in the sagittal plane, shows that the tibia experiences a lateral shift during the early phase of flexion. This can be attributed to the shape of the medial femoral condyle being larger than the lateral femoral condyle. Hence, when the knee joint unlocks to begin flexion, the tibia rotates internally relative to the femur about the medial femoral condyle. This would also suggest that the tibial rotation axis lies along the medial compartment of the knee.

None of the techniques used for total knee replacement restored the kinematics of the knee to the intact state. The femur in the reconstructed knees rolled excessively on the tibia before gliding in the range in which the ACL is most effective in limiting anterior

tibial translation. The disintegration of the rolling-gliding movement of the knee was exacerbated. Total knee replacement using the PCL retaining technique paralleled the knee in the intact state closer than the total knee replacement using the PCL sacrificing technique. Beyond 70 degrees of flexion, the total knee replacement using the PCL sacrificing technique moved posteriorly relative to the intact state and the total knee replacement using the PCL retaining technique. This behavior is consistent with the fact that the PCL is most effective in limiting posterior tibial translation in the range of 70-90 degrees of flexion.

VI. RECOMMENDATIONS

The most glaring weakness in this six degree of freedom model is the 2 degrees per second rate at which the knee specimens flex and extend. This rate is controlled by the speed at which the data acquisition system can record the data. Studies show that the cruciate ligaments when loaded exhibit nonlinear load-displacement plots, i.e., the compliance of the cruciate ligaments depend on the rate at which their loaded. Hence, future studies will attempt to record data that more accurately represents natural knee flexion and extension. This will require a data acquisition system with a greater bandwidth and transducers with smaller measurement uncertainties. The non-orthogonality of the human knee is such that the behavior of the kinematics depends on where the reference point is chosen. In the present model, the reference point is the point on the surface of the medial femoral condyle along the femoral transepicondylar pin. Future studies will incorporate a second set of translational transducers on the lateral femoral condyle. The difference in the motion along the flexion axis will then represent relative displacement at the knee joint. In this way, the gross translation of the knee joint can be more accurately estimated.

Future studies will also constrain various degrees of freedom in an effort to compare the kinematics that have been published using planar models as well as increase loading conditions to more closely simulate walking and squatting.

Finally, future tests will synchronize the sequence of events between data acquisition and test commencement to ensure the starting point of one half-cycle run begins where the other half-cycle terminated. This will eliminate the apparent discontinuity between the two end points observed in the plots of the hysteresis loops.

GLOSSARY OF TERMS

ABDUCTION - Move laterally away from the mid-line

ADDUCTION - Move medially toward the mid-line

ANTERIOR CRUCIATE LIGAMENT - Crossed shaped ligament that limits anterior tibial translation

ANTERIOR-POSTERIOR DRAWER - Horizontal motion in the sagittal plane

ARTHROPLASTY - Surgical procedure that involves inserting a prosthesis into the knee joint

ARTHROSCOPY - Surgical procedure used by an orthopedist to look into a joint using a specially designed, lighted, hollow instrument

CAPSULE - Covering or sheathing

COMPRESSION-DISTRACTION - Movement along the tibiofemoral axis

CONDYLE - Bulbous shaped part of the femur at the knee joint

CRUCIATE - Cross shaped

EXTENSION - Straightening or stretching of a limb which is contracted

FEMUR - Long bone in the upper part of the leg connecting the hip to the knee

FLEXION - Relative rotation at a joint in the sagittal plane

GASTROSOLEUS - Main calf muscle in rear portion of lower leg

HAMSTRINGS- Muscle group located to the rear of the thigh

LIGAMENT - Tough connective tissue holding bones together

MEDULLARY - Canal in the center of the leg bones

MENISCUS - Crescent shaped cartilage lying on tibia plateau within the tibiofemoral joint

PATELLA - Knee-cap

PATHOKINEMATICS - Dealing with the nature of the kinematics

POSTERIOR CRUCIATE LIGAMENT - Cross shaped ligament that limits posterior tibial translation

QUADRICEPS FEMORIS - Large muscle group in front of thigh

SCREW-HOME MECHANISM - Manner in which the tibiofemoral joint locks itself upon full extension

TIBIA - Weight bearing long bone in the lower leg

TIBIOFEMORAL - Pertaining to the femur and the tibia

TRANSEPICONDYLAR - Along the medial-lateral axis of the femoral condyles

VALGUS - Turning out of the foot away from the mid-line

VARUS - Turning in of the foot toward the mid-line

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